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# Low-complexity intrauterine pressure estimation using the Teager energy operator on electrohysterographic recordings

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## Abstract

Monitoring the progression of maternal uterine activity provides important prognostic information during pregnancy and parturition. Currently used methods for intrauterine pressure (IUP) measurement are unsuitable for long-term observation of uterine activity. The abdominally measured electrohysterogram (EHG) provides a non-invasive alternative to the existing methods for long-term ambulatory uterine contraction monitoring. A new low-complexity method for IUP estimation based on the Teager energy (TE) operator is proposed. The TE operator was used as it mimics the physiologic phenomena underlying the generation of the EHG signals. Several EHG signal analysis methods for IUP estimation from the literature are compared with the new TE method. The comparison is based on correlation and root mean square error of the IUP estimate with the gold standard internally measured IUP as well as their respective computational complexity. The proposed method results in a superior IUP estimation accuracy and complexity compared to state-of-the-art methods from the literature, with a complexity as much as 55 times lower.

Therefore, the proposed method offers a valuable new option for long-term ambulatory uterine monitoring.

Keywords: pregnancy monitoring, electrohysterography, intrauterine pressure (IUP), Teager energy, low-complexity

## 1. Introduction

Preterm birth is a major problem in obstetrics and is associated with over 75% perinatal mortality and more than 50% perinatal and long-term morbidity (Goldberg and McClure 2010). Early prediction of the onset of preterm delivery by accurate monitoring of patients at risk is crucial for prevention of preterm birth, as it allows for timely intervention by the physician. Interventions such as administration of tocolytic drugs or progesterone supplements, or advising the patient to rest, can in return significantly reduce preterm birth rates (Flood and Malone 2012). Tracking the progression of uterine activity during pregnancy, by determination of frequency, duration, and amplitude of the uterine contractions, gives insight in the time to delivery and can be used to predict preterm delivery (Bentley *et al* 1990). Accurate measurement of the intrauterine pressure (IUP), which increases as a result of increased uterine activity, is required for correct calculation of these variables. To reliably track the progression of uterine activity for prevention of preterm birth, long term measurement of the IUP throughout the last months of pregnancy is essential.

In clinical practice, uterine activity is generally monitored by direct or indirect measurement of the IUP during relatively short time intervals. External tocography is currently the most widely used method to monitor uterine activity during pregnancy and delivery. It uses a tocodynamometer, which consists of a strain gauge transducer placed on the abdomen to indirectly estimate the IUP by measuring the deformation of the abdominal surface. External tocography can be used throughout pregnancy, as it is non-invasive. However, deriving the IUP estimate from an indirect mechanical measure makes it very susceptible to movement artifacts and unreliable in obese patients (Eswaran *et al* 2002, Euliano *et al* 2007). This can result in a low sensitivity, affecting the accuracy of the estimated amplitude, duration, and frequency of contractions, influencing estimation of the fetal health. Patients are therefore required to remain steady during measurements, limiting the use of tocodynamometry for long-term ambulatory measurements.

During delivery, an internal uterine pressure catheter (IUPC) can be used to obtain a quantitative direct measurement of the IUP in the amniotic fluid. The obtained IUP signal is very accurate and currently used as the gold-standard measurement. However, the use of an IUPC requires rupturing of the amniotic membranes, increasing the risk of infections or damage to the fetus; it is therefore only employed during parturition in complicated deliveries which require labor induction or augmentation (Garfield *et al* 2002).

An alternative method for uterine activity monitoring is based on the electrohysterogram (EHG), which is the measurement of the bioelectrical signals resulting from propagation of action potentials in the myometrium, i.e. the uterine muscle. The EHG gives an indication of the uterine activity, because propagation of action potentials through an adequate number of cells results in the contraction of the uterine muscle and, consequently, in an increase in IUP. The IUP increase associated with each contraction can, therefore, be estimated, providing essential information on the uterine activity (Garfield and Maner 2007, Euliano *et al* 2013). During pregnancy, the electrical resistance between the myometrial cells is relatively high, while at term low-resistance paths form, leading to the occurrence of effective contractions

(Garfield *et al* 1998). Therefore, the EHG can possibly provide an indication of the time to delivery (de Lau *et al* 2013). Because this technique is non-invasive, due to the use of abdominal electrodes, it is suitable for long-term uterine observation throughout pregnancy aimed at prevention of preterm birth.

Various methods have previously been proposed for EHG analysis, comprising statistical approaches (Horoba *et al* 1999), filtering techniques (Ramondt *et al* 1986, Horoba *et al* 2001), fast Fourier transform (FFT) and wavelet transform based methods (Buhimschi *et al* 1997, Eswaran *et al* 2002, Diab *et al* 2012b), as well as evaluation of nonlinear connectivity (Diab *et al* 2012a). However, only recent studies focused on EHG analysis as an alternative method for quantitative IUP estimation (Jezewski *et al* 2005, Skowronski *et al* 2006, Rabotti *et al* 2008).

Jezewski *et al* estimated the contraction pattern by taking the root mean square (RMS) value of the EHG and compared it to the simultaneously recorded external tocogram (Jezewski *et al* 2005). The estimated contraction pattern showed a high correlation with the externally measured tocogram, and was hence shown to be a reliable estimator for the contraction frequency. Skowronski *et al* proposed optimal linear filtering by calculating the magnitude of the EHG signal after low-pass filtering, followed by a Wiener filter to estimate the IUP (Skowronski *et al* 2006). The method showed a good resemblance to the simultaneously measured IUPC signal. However, as the Wiener filter requires a 10 min initialization period, in which the IUPC signal is measured as a reference, the method is unsuitable for long-term ambulatory measurements. Rabotti *et al* obtained an IUP estimate employing a spectrogram-based technique, which showed a high correlation with the simultaneously measured IUPC signal (Rabotti *et al* 2008). Compared to other algorithms from the literature, a superior estimation accuracy was obtained by mimicking the physiologic phenomena underlying the generation and propagation of the EHG signals. The EHG amplitude is proportional to the number of muscle cells involved in the contraction, while the propagation of action potentials through the uterus is reflected in the EHG spectrum. Using frequency compensation by scaling all spectrogram frequency bands accounts for both physiologic effects (Rabotti *et al* 2008). In addition a patient specific second-order model was used to reduce the influence of muscle fatigue and physiology related differences in electrical conductivity and contractile efficiency. These improvements, however, increased the complexity of the method accordingly, making it unsuitable for long-term ambulatory uterine activity monitoring.

With the aim of long term ambulatory pregnancy monitoring, a low complexity method for EHG-based IUP estimation was developed using the Teager energy (TE) operator, mimicking the frequency compensation used in the spectrogram method by Rabotti *et al*. The short support width of the TE operator results in a low computational complexity, while a high-quality IUP estimate is obtained because of the generated frequency-weighted energy estimate. The algorithm is validated by comparing the results with those of various state-of-the-art algorithms from the literature based on IUP estimation accuracy and computational complexity. Like Skowronski *et al* and Rabotti *et al*, the algorithm accuracy is determined by comparing all IUP estimates with the simultaneously measured IUPC signals. The IUPC was chosen as the reference measurement method over the external tocogram as it provides the most accurate IUP measurements. The estimated IUP can directly be compared to the IUPC signal based on shape similarity, which allows for reliable determination of relative contraction amplitude, length and frequency. Results show that the proposed algorithm has both an improved IUP estimation accuracy and reduced computational complexity compared to the algorithms from the literature.

Section 2 first describes the state-of-the-art IUP estimation algorithms from the literature. Next, the TE operator is introduced, followed by the introduction of a novel TE-based IUP

estimation algorithm. Section 3 presents the dataset of abdominal EHG measurements with simultaneous IUPC recordings, which was used for validation. The validation methodology, based on comparison of both estimation accuracy and computational complexity, is also presented. Section 4 shows a comparison of the accuracy and complexity results for all algorithms, followed by a discussion and conclusion in section 5.

## 2. Methodology

### 2.1. IUP estimation methods from the literature

**2.1.1. Spectrogram-based IUP estimation.** Rabotti *et al* proposed the use of the spectrogram to calculate the frequency-weighted energy of the EHG signal as an estimate of the IUP (Rabotti *et al* 2008). First, the time-frequency representation of the EHG signal  $\rho[n, f]$  is calculated based on the squared magnitude of the short-time Fourier transform (STFT) with a Hamming window  $w[m]$  of length 70 s as defined by

$$\rho[n, f] = \left| \sum_{m=-\infty}^{+\infty} x[m]w[m-n]e^{-j2\pi fm} \right|^2. \quad (1)$$

The unnormalized first statistical moment  $\Psi[n]$  is calculated by scaling  $\rho[n, f]$  for each frequency interval in the EHG frequency range [ $f_{\min} = 0.3$  Hz,  $f_{\max} = 0.8$  Hz] by its mean frequency  $f$ , as described by

$$\Psi[n] = \sum_{f_{\min}}^{f_{\max}} f \cdot \rho[n, f]. \quad (2)$$

The frequency intervals outside the EHG frequency range are multiplied by zero, which results in a band-pass filtering effect. Therefore,  $\Psi[n]$  uses the simultaneous increase of both the frequency- and amplitude-related features of the EHG signal to determine the IUP wave estimate.

In Rabotti *et al* (2008), a second-order model is used to reduce the possible influence of muscle fatigue and the physiology underlying the relation between electrical activity and contractility. Use of the model only results in a minor improvement in IUP estimation accuracy and requires estimation of the patient-specific coefficients; therefore, here  $\Psi[m]$  is used as the spectrogram-based IUP estimate  $IUP_s$ .

**2.1.2. RMS-based IUP estimation.** Jezewski *et al* proposed a RMS-based IUP estimation algorithm, where the EHG is first band-pass filtered between  $f_{\min} = 0.05$  Hz and  $f_{\max} = 5$  Hz, resulting in the filtered signal  $x_f[n]$  (Jezewski *et al* 2005). Next, an initial IUP estimate  $\widehat{IUP}_{RMS}$  is calculated as

$$\widehat{IUP}_{RMS}[n] = \left[ \frac{\sum_{m=-M/2}^{+M/2} x_f[m+n]^2 \cdot w[m]}{\sum_{m=-M/2}^{+M/2} w[m]} \right]^{\frac{1}{2}}, \quad (3)$$

where  $w[m]$  is a Hamming window, which is used to smoothen the estimate and reduce edge effects,  $M \equiv 60$  s is the length of the Hamming window, and  $n$  is incremented in 3-s steps. Therefore, the resulting signal  $\widehat{IUP}_{RMS}[n]$  has a sampling frequency of 1/3 Hz. Finally,  $\widehat{IUP}_{RMS}[n]$  is resampled to the original sampling frequency, yielding the RMS IUP estimate  $IUP_{RMS}[n]$ .

**2.1.3. Absolute-value-based IUP estimation.** Skowronski *et al* proposed a method for IUP estimation based on rectification of the EHG signal (Skowronski *et al* 2006), where the EHG signal was first normalized to have zero mean and unity variance. This first step is not feasible in a real-time ambulatory system, as the entire signal segment has to be available; therefore, normalization is omitted in our implementation. Hence, the EHG signal is first low-pass filtered with a cut-off frequency of  $f_{\max} = 2$  Hz and down-sampled to a sampling frequency of 4 Hz. Consecutively, the EHG signal is rectified to obtain a first approximation of the energy in the EHG signal. After low-pass filtering with a cut-off frequency of 0.1 Hz to smoothen the EHG energy signal, the signal is further downsampled to 0.2 Hz, resulting in the absolute-value-based IUP estimate  $IUP_{\text{ABS}}$ .

In Skowronski *et al* (2006), a Wiener filter  $W(n)$  was used to better predict the IUP from the EHG signal. The Wiener filter coefficients were obtained by reducing the error signal  $e(n) = d(n) - IUP_{\text{ABS}}(n) * W(n)$  in 10 min signal segments for each patient in a mean-squared sense. Here,  $d(n)$  is the reference IUPC signal and  $*$  is the convolution operator. As no IUPC reference signal is available in real ambulatory measurements, this extension of the method is unsuitable for long-term ambulatory measurements and, therefore, the Wiener filter was not included in our implementation of the algorithm.

## 2.2. Teager energy

The total energy of a system is the sum of the stored potential energy and kinetic energy. Therefore, the energy required to give a mass  $m$  a harmonic motion is given by

$$E = \frac{1}{2}m \cdot \omega^2 A^2 \quad (4)$$

where  $\omega$  and  $A$  are the frequency and amplitude of oscillation, respectively. The TE operator  $\Phi$  introduced by Kaiser can be used to calculate the instantaneous energy of such a harmonic signal (Kaiser 1990, 1993), resulting in an energy proportional to the product of frequency squared and amplitude squared. In the continuous time domain, the TE operator  $\Phi(t)$  is defined as a function of the input signal  $x(t)$  by

$$\Phi(t) = \left( \frac{dx(t)}{dt} \right)^2 - x(t) \frac{d^2x(t)}{dt^2}, \quad (5)$$

from which it is clear that the TE represents a property of the signal depending only on the signal and its first two time derivatives. By substituting  $x(t)$  by a simple oscillatory signal defined by

$$x(t) = A \cos(\omega t + \phi), \quad (6)$$

where  $\phi$  is an initial phase, equation (5) can be rewritten to

$$\Phi(t) = \omega^2 A^2 \propto E = \frac{1}{2}m \cdot \omega^2 A^2, \quad (7)$$

showing that  $\Phi(t)$  can be used to estimate  $E$ . In the discrete time domain, the application of the TE operator  $\Phi$  to an input signal  $x[n]$  can be defined as

$$\Phi[n] = x[n]^2 - x[n+1]x[n-1], \quad (8)$$

where  $n$  indicates the index of the sample in the signal  $x$ , from which it can be observed that  $\Phi[n]$  only spans three consecutive samples of  $x$  to calculate the instantaneous energy at time  $n$ .

The short support width of  $\Phi[n]$  results in an excellent time resolution when calculating the instantaneous energy. The TE operator is also robust to white noise and invariant to tonal interference (Dunn *et al* 1993, Sherman *et al* 1997). However, due to the short support width, the TE operator is sensitive to transient signals and hence susceptible to transient noise.

Additionally, the TE operator is not linear when superimposing two or more signals, which results in a misestimation of the signal energy (Kaiser 1990). Therefore, in order to calculate an exact energy using the TE operator, the various frequency components of the analyzed signals need to be separated prior to energy calculation. To this end, a multiband solution for energy tracking over a wideband signal is introduced in Evangelopoulos and Maragos (2006), which requires the use of a filter bank followed by calculation of the TE in each sub-band.

### 2.3. TE-based IUP estimation

Myometrial cells can generate action potentials or can be excited by action potentials generated by a neighboring cell. Uterine contractions are the result of this generation and propagation of electrical activity in the form of action potentials through myometrial cells. The propagation of the action potentials is reflected in the spectral properties of the EHG signal, while the number of cells involved in the contraction is proportional to the amplitude of the EHG signal. As described in section 2.2, the TE operator is proportional to the square of both signal amplitude and frequency and, therefore, accounts for the contribution of both the number of cells (amplitude) and the propagation of action potentials (frequency). This way, the TE provides a representation of the physiological process leading to contraction of the myometrium and results in the frequency-weighted energy of the EHG signal.

To reduce the influence of respiration artifacts and morphological changes of the ECG, a clean EHG signal is obtained by band-pass filtering, with the low and high cut-off frequencies  $f_{\min} = 0.3$  Hz and  $f_{\max} = 0.8$  Hz, respectively, in agreement with previous studies (Buhimschi *et al* 1997, Rabotti *et al* 2008, Maner *et al* 2003, Leman *et al* 1999). To reduce the computational complexity, the input signal is first down-sampled to 4 Hz after low-pass filtering at 2 Hz. Next, the down-sampled signal is band-pass filtered, providing the filtered EHG signal  $x_f[n]$ . The filtered signal  $x_f[n]$  is processed using the TE operator in (8) to yield the instantaneous energy of the EHG signal. Finally, the absolute value of the moving average of  $\Phi[n]$  is calculated over a 30 s interval, after which the square root is taken to obtain the TE IUP estimate ( $IUP_{TE}$ ), as defined by

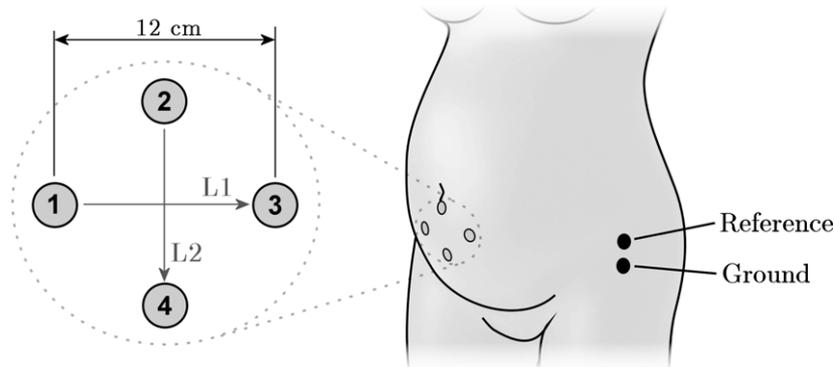
$$IUP_{TE}[n] = \left| \frac{1}{M} \sum_{m=-M/2}^{+M/2} \Phi[x_f[m+n]] \right|^{\frac{1}{2}}, \quad (9)$$

with,  $M \equiv 30$ s. The absolute value is used as the moving average of  $\Phi[n]$  can theoretically become negative under the influence of interference (Dunn *et al* 1993). Because the misestimation under the influence of interference is on average zero and the averaging interval is relatively long, it is very unlikely to occur. The resulting  $IUP_{TE}$  provides a simplified reproduction of the physiology-based frequency-weighted energy estimate proposed by Rabotti *et al* (2008). However, according to standard EMG force estimation and different from Rabotti *et al* (2008), a square root is used to retain the original measurement units (Merletti and Parker 2004). Furthermore, because the energy of the EHG is concentrated in a limited frequency band, the use of a single TE operator was deemed suitable, eliminating the need for the use of a filter bank and multiple TE transforms.

## 3. Validation

### 3.1. Measurements

The same dataset as described in Rabotti *et al* (2008) was used for validation of the proposed algorithm. This dataset contains 21 measurements on nine women during labor. All women



**Figure 1.** Electrode configuration used for IUP measurements consisting of 4 electrodes from which the two bipolar leads L1 and L2 are derived.

signed an informed consent. Recordings were made using four active unipolar electrodes in a diamond shape as well as a ground and reference electrode, placed on the abdomen as shown in figure 1 (Rabotti *et al* 2008). The measured signals were bipolarized to reduce a large portion of the noise affecting the EHG, the most important of which are the maternal electrocardiogram, part of the movement artifacts, and electromagnetic noise (Graczyk *et al* 1995). Two bipolar leads *L1* and *L2* were obtained from the recorded signals in a horizontal and vertical direction, respectively. All electrodes, which were contact Ag–AgCl electrodes, were placed on the abdomen after skin preparation with abrasive paste to reduce skin impedance. The IUP was measured simultaneously by a Koala M1333A (Philips Medical Systems, Best, The Netherlands) IUPC inserted in the uterine cavity, which was applied due to medical prescription. The EHG and IUP were both recorded at 1 kHz with a 20-bit discretization resolution using an M-PAQ (Maastricht Instruments B.V., The Netherlands). A subset of 17 recordings from seven patients was selected based on the requirements that the reference IUPC signal should show a clear IUP and the EHG lead should not show any significant measurement artifacts, e.g. due to broken wires or bad contacts. All recordings from a single patient were considered as a single measurement, resulting in seven measurements ranging from 26 to 138 min with a total length of 8 hours and 42 min.

Both the EHG and IUP signal were preprocessed before use in the comparative analysis. The IUP signal, which is adopted as the golden standard reference signal, was cleaned to minimize spikes caused by movement artifacts using a non-causal centered median filter with a length of 5 s (Jezewski *et al* 2005). Both bipolar EHG signals as well as the IUPC signal were downsampled to 20 Hz using an anti-aliasing filter at 5 Hz. This is possible because the all EHG signals are present in the 0.3 Hz to 3.0 Hz frequency band (Buhimschi *et al* 1997, Rabotti *et al* 2008, Maner *et al* 2003, Leman *et al* 1999).

### 3.2. Quality measures

Validation of the various IUP estimation algorithms was performed by comparing both the accuracy of the estimated IUP as well as their complexity.

**3.2.1. Accuracy measures.** Two different accuracy measures were used to determine the accuracy of the various algorithms at estimating the IUP. The correlation coefficient  $r$ , which gives a similarity in shape between the estimated and reference IUP waveforms, is defined as

$$r = \frac{\sum_{i=1}^N (x_i - \bar{x})(y_i - \bar{y})}{\left[ \sum_{i=1}^N (x_i - \bar{x})^2 \sum_{i=1}^N (y_i - \bar{y})^2 \right]^{\frac{1}{2}}}, \quad (10)$$

with  $x$  being the IUP estimate,  $y$  the IUPC signal,  $\bar{x}$  and  $\bar{y}$  their means, respectively, and  $N$  the total number of samples in the signal. The correlation coefficient  $r$  is in the range  $-1 \leq r \leq 1$ , where a higher value defines a better similarity in shape between the two waveforms. The root-mean-square error (RMSE), which is defined by

$$\text{RMSE} = \left[ \frac{\sum_{i=1}^N (y_i - x_i)^2}{N} \right]^{\frac{1}{2}}, \quad (11)$$

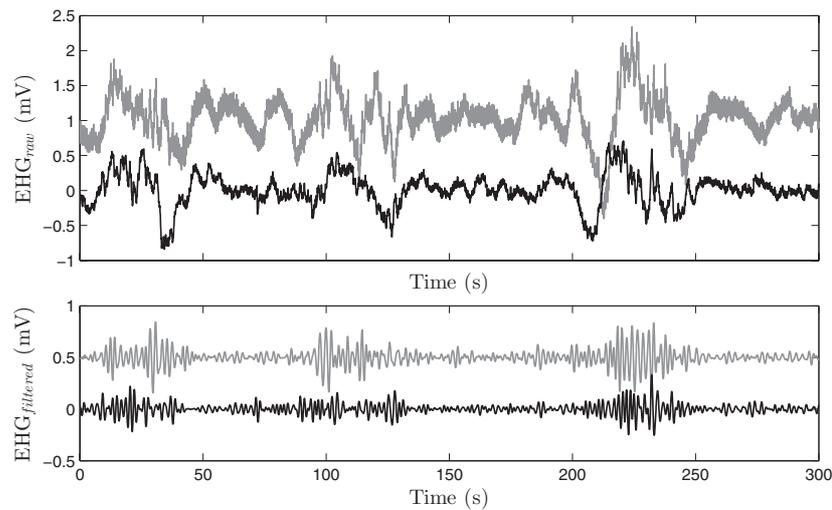
on the other hand, not only depends on similarity in shape, but is also influenced by scaling factors and baseline offset<sup>3</sup>. To reduce the influence of these effects, the basal tones of the IUP estimates for all patients  $P$  were removed using the method described in Jezewski *et al* (2005) and their amplitudes were scaled by a factor  $A$  to best match the IUPC signal. Additionally, as the uterine pressure is a direct result of action potentials spreading through the myometrium, the IUP increase is always temporally delayed with respect to the electrical activation. Therefore, the IUP estimate of each patient was delayed by  $D$  samples to mimic the delayed pressure increase as a result of the increased bioelectrical activity in the uterus. A cross-validation method is used to find fixed values for  $A$  and  $D$  and assess how the algorithms accuracy generalizes to an independent dataset. Here,  $A_p$  and  $D_p$  are estimated for each patient  $p \in P = \{1, \dots, 7\}$  as the mean value of the optimized estimates for all the other patients  $\{P \setminus p\}$ . The resulting parameters  $A_p$  and  $D_p$  are used to calculate the scaled and shifted IUP estimates, which are used for calculation of  $r$  and  $\text{RMSE}$ , as described above.

**3.2.2. Complexity measure.** The average number of multiplications per sample (MPS) is used as a measure of computational complexity. All operations with a complexity higher than a multiplication, e.g. a division or square root, can be represented by multiple multiplications. Both operations have a complexity in the order of  $O[n]$ , where  $n$  is the accuracy of a value in number of bits. Assuming an accuracy of 16 bits for both numerator and denominator, a division and square root are substituted by 9 and 17 multiplications, respectively (Robison 2005). All simple operations, e.g. addition, subtraction, or bit-shift, are estimated but neglected.

Like the simple operations, the computational complexity of the STFT-based spectrogram can also be expressed as the number of multiplications used. The STFT can be represented as a sliding fixed-length window for which at each position an FFT is calculated. An efficient way of implementing the FFT uses the butterfly principle. This is a recursive method combining partial FFT results at a higher scale which restricts the length of the used window to powers of 2. Based on the butterfly principle the STFT can be implemented with a complexity of  $O[N \log_2(N)]$  (Rabiner and Schafer 1971), where  $N$  is the length of the STFT window. Further optimizations have led to a number of multiplications in the order of  $O[(N/2) \log_2(N)]$  (Rabiner and Gold 1975, Farhang-Boroujeny and Lim 1992). The spectrogram method by Rabotti *et al* employs a window of 1400 samples. This can only be efficiently implemented by extending it to a length equal to a power of 2 by adding zeros. This results in a window length  $N = 2048$  samples, requiring 8194 multiplications to obtain the STFT for a single position of the sliding window (Vetterli and Nussbaumer 1984).

For a fair comparison of the algorithms presented in section 2, all algorithms are optimized for reduced computational complexity using basic optimization methods.

<sup>3</sup> It can be noted that the RMSE retains the same units of pressure as the reference IUP signal (mmHg).



**Figure 2.** Both subplots show examples of EHG signals over a period of 5 min from the horizontal (solid black) and vertical (solid gray) bipolar leads. The top and bottom plots show the raw EHG signal after 5 Hz low-pass filtering to reduce aliasing and after filtering between 0.3 and 0.8 Hz, respectively.

- Sliding windows are implemented such that all multiplications by a single value are only performed once. In case of a symmetric window, this typically reduces the number of multiplications by a factor 2.
- Averaging filters are implemented by means of a sliding cumulative sum, requiring a single addition, subtraction, and division for each shift.
- Multiplications and divisions by a power of 2 are replaced by a bit-shift operation.
- Divisions by a constant  $K$  are replaced by a multiplication with a precalculated value  $\hat{K} = 1/K$ .

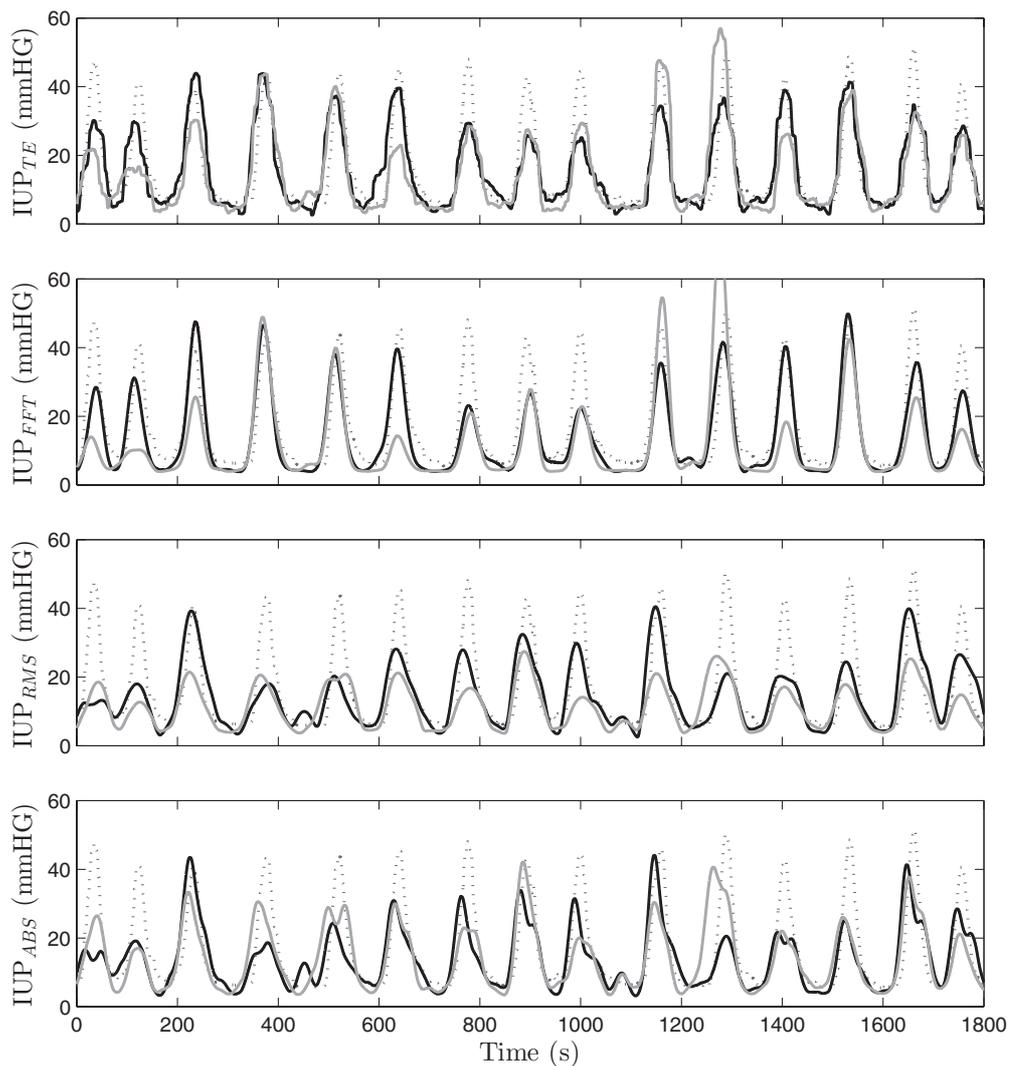
Additionally, all filters were implemented as finite impulse response filters, where the filter order was tuned such that the cut-off steepness of the filter is  $80 \text{ dB dec}^{-1}$ .

#### 4. Results

Figure 2 shows an example of the EHG signal over a period of 5 min for both the horizontal and vertical bipolar measurement channels both before and after filtering. In the filtered EHG signal, we can clearly see three periods with increased signal amplitude, each of which coincides with a contraction.

Figure 3 shows a comparison of the four IUP estimates over a 30 min interval, obtained using all the methods described in this paper. A clear resemblance between the  $IUP_{TE}$  and  $IUP_{FFT}$  subplots can be observed due to the use of a frequency-weighted IUP estimate. The slightly jagged outline of the  $IUP_{TE}$  plot is due to the use of the simple square window.

Table 1 shows both the correlation coefficient  $r$  and RMSE for each of the described IUP estimation methods as well as their standard deviation, for the horizontal (L1) and vertical (L2) channel, respectively. The correlation coefficient  $r$  of the proposed TE method shows a significant improvement ( $p < 0.01$ ) over the RMS and ABS methods for lead L1. For the vertical lead (L2) the improvement in  $r$  of the proposed method over those from the literature



**Figure 3.** All four subplots show the IUP estimated from the horizontal (solid black) and vertical (solid gray) bipolar EHG signals together with the IUPC signal (dotted). From top to bottom the IUP estimates were calculated using the TE, spectrogram, RMS, and ABS methods, respectively.

is non-significant, as both RMS and ABS methods perform markedly better for L2 compared to L1.

Table 2 shows the correlation coefficient  $r$  and RMSE for each of the described methods when using the mean of the IUP estimates from both lead L1 and L2 as the final IUP estimate; a significant improvement in  $r$  for the proposed TE method over the RMS and ABS methods is achieved.

Table 3 shows the total number of operations needed for each of the four described algorithms when operating on an EHG signal sampled at 20 Hz. The four columns from left to right give the number of square roots, divisions, multiplications, and summations, respectively. Although an integer number of operations is used, the number of operations per sample can be a non-integer number due to sample-rate conversions.

**Table 1.** Accuracy comparison of IUP estimates obtained using the described algorithms using lead L1 (horizontal) and lead L2 (vertical), respectively.

		<i>r</i>	RMSE (mmHg)
L1	<b>Teager energy</b>	<b>0.68 ± 0.12</b>	<b>13.9 ± 2.8</b>
	Spectrogram (Rabotti <i>et al</i> 2008)	0.62 ± 0.18	19.4 ± 9.6
	RMS (Jezewski <i>et al</i> 2005)	0.48 ± 0.30	15.7 ± 3.2
	ABS (Skowronski <i>et al</i> 2006)	0.43 ± 0.27	16.5 ± 3.8
L2	<b>Teager energy</b>	<b>0.68 ± 0.25</b>	<b>20.4 ± 8.8</b>
	Spectrogram (Rabotti <i>et al</i> 2008)	0.61 ± 0.29	39.8 ± 33.3
	RMS (Jezewski <i>et al</i> 2005)	0.58 ± 0.29	17.6 ± 13.8
	ABS (Skowronski <i>et al</i> 2006)	0.56 ± 0.28	17.3 ± 12.1

**Table 2.** Performance comparison of IUP estimation algorithms using a one-to-one mixture of both IUP estimations.

		<i>r</i>	RMSE (mmHg)
<b>Teager energy</b>		<b>0.74 ± 0.13</b>	<b>16.0 ± 4.1</b>
	Spectrogram (Rabotti <i>et al</i> 2008)	0.63 ± 0.23	30.7 ± 18.9
	RMS (Jezewski <i>et al</i> 2005)	0.55 ± 0.31	17.7 ± 10.3
	ABS (Skowronski <i>et al</i> 2006)	0.52 ± 0.29	18.3 ± 8.4

**Table 3.** Number of computations per sample needed to perform the various operations for each of the algorithms, based on a 20 Hz sampling frequency.

		$\sqrt{x}$	$x/y$	$x \cdot y$	$x \pm y$
<b>Teager energy</b>	LP filter [−2.0]	0	0	79	78
	BP filter [0.34–0.8] <sup>a</sup>	0	0	66	65
	TE operator <sup>a</sup>	0	0	0.4	0.2
	Mean <sup>a</sup>	0	0	0.2	0.4
	Square root <sup>a</sup>	0.2	0	0.2	0.4
<b>Total</b>		<b>0.2</b>	<b>0</b>	<b>145.8</b>	<b>144</b>
Spectrogram	STFT	0	0	8194	28 676
	Square	0	0	1	0
	Freq. comp.	0	0	56	55
<b>Total</b>		<b>0</b>	<b>0</b>	<b>8251</b>	<b>28 731</b>
RMS	BP filter [0.05–5.0]	0	0	582	581
	Hanning window	0	0	600	1199
	RMS	1	1	1	2
<b>Total</b>		<b>1</b>	<b>1</b>	<b>1183</b>	<b>1802</b>
ABS	LP filter [−2.0]	0	0	79	78
	modulus <sup>a</sup>	0	0	0	0.2
	LP filter [−0.1] <sup>a</sup>	0	0	62	61.8
	Wiener filter <sup>b</sup>	0	0	0.1	0.1
<b>Total</b>		<b>0</b>	<b>0</b>	<b>141.1</b>	<b>140.1</b>

<sup>a</sup> Calculation based on 4 Hz local sample frequency.

<sup>b</sup> Calculation based on 0.2 Hz local sample frequency.

### 5. Discussion and conclusion

In this paper, we propose a non-invasive method to estimate the IUP based on the assessment of the energy in the EHG signal using the TE operator, with the ultimate goal of long-term ambulatory uterine monitoring. The method was tested on a set of 17 measurements from

seven women in labor. Each measurement contains four monopolar EHG signals and a simultaneously measured IUPC signal, each sampled at 1 kHz. The TE algorithm was compared with three alternative IUP estimation methods proposed in the literature in terms of both IUP estimation accuracy and computational complexity. These methods are the spectrogram method proposed in Rabotti *et al* (2008), the RMS method described in Jezewski *et al* (2005), and the ABS method described in Skowronski *et al* (2006).

The proposed method reflects the fundamental physiologic phenomena underlying the generation of the EHG signal by using the TE operator as a basis for the IUP estimate. This implies that the influence of the signal frequency content on the contractile strength is taken into account (Rabotti *et al* 2008). Similar to the methods that are commonly used in EMG analysis, the original units of measurement are retained by application of the square root (Jezewski *et al* 2005, Skowronski *et al* 2006, Merletti and Parker 2004). All together, this results in a superior accuracy of the IUP estimates, with a correlation coefficient  $r$  of 0.68 for a single bipolar channel, and 0.74 when combining two channel estimates. The high estimation accuracy compares favorably to that of the spectrogram method, which results in a correlation coefficient  $r$  of 0.62. In case no square-root operation is performed the TE method yields similar results to the spectrogram method, showing that retaining the original units of measurement, as proposed for EMG analysis, is advantageous. The IUP estimates obtained using the RMS and ABS analyses show a lower correlation with the reference IUPC signal ( $p < 0.01$ ).

It can be noted that there is a clear difference in the correlation coefficient between the horizontal (L1) and vertical (L2) bipolar leads for both the RMS and the ABS methods. This difference arises because the measured frequency content depends on the relative orientation between the bipolar leads and the propagation direction of action potentials in the uterus. Additionally, a change in measurement orientation with respect to the various artifact sources, e.g. striated abdominal muscles, movement due to respiration, and the fetal and maternal heart, results in a different artifact amplitude. Differences in the filtering interval between the proposed method and those from the literature, as well as the application of frequency weighing, makes the proposed method more robust to these artifacts and the changes in measurement orientation. Hence, compared to the other methods, the proposed TE method provides a more reliable IUP estimator and, therefore, making it the most suitable for use in an ambulatory setting.

The RMSE obtained by the proposed method is, on average, only slightly better than that of the RMS and ABS methods. The limited improvement in RMSE is because the inter-patient variability, which has a major influence on amplitude scaling, is relatively large, influencing the RMSE results. Due to the quadratic nature of the spectrogram-based estimate, the amplitude variation introduced by inter-patient differences is amplified, resulting in a significantly higher RMSE compared to that of the proposed algorithm. The second-order model used in Rabotti *et al* (2008) reduces this effect, resulting in a significant improvement in RMSE when no patient-specific optimizations are performed. In case of patient specific amplitude scaling, the RMSE of all algorithms reduces to the levels given in Rabotti *et al* (2008) with an RMSE of  $10.0 \pm 3.83$  mmHg for the proposed TE algorithm, equivalent to the IUPC signal accuracy reported in the literature (Dowdle 2003, Arulkumaran *et al* 1991).

Evidently, physiological inter-patient differences result in large absolute errors in IUP amplitude. However, none of the clinically relevant parameters depend on the exact IUP amplitude; instead, clinical IUP evaluation relies mainly on the timing and shape of the IUP estimate, especially when monitoring prior to labor (Euliano *et al* 2013). The IUPC was used to obtain a golden standard signal from which the clinically relevant parameters can easily be extracted and used for comparison and validation. The correlation coefficient defines the estimation accuracy based on similarity in shape and timing without considering amplitude

scaling. Therefore, in view of our goal of long-term ambulatory uterine contraction monitoring, the correlation coefficient is the most relevant of the two proposed accuracy metrics.

Because of the short support width of the TE operator, the total computational complexity of the proposed algorithm is very low, with a complexity equivalent to 145.8 MPS based on a 20 Hz sampling frequency. This amounts to an improvement by a factor 55 and 8 compared to the spectrogram and RMS methods, which use 8251 MPS and 1208 MPS, respectively. Only the ABS method, which has a computational complexity equal to 141.1 MPS based on a 20 Hz sampling frequency, has a complexity comparable to the proposed TE method. Like the proposed method, it also uses sample rate reduction to lower the complexity. Therefore, both the proposed algorithm and the ABS method are suitable for real-time applications. Due to the low computational complexity as well as the significantly higher correlation between IUP estimate and IUPC signal, the proposed method is suitable for long-term ambulatory monitoring of the uterine activity during pregnancy. This method might therefore provide an important possibility for timely detection of preterm delivery, which requires long-term uterine activity measurements.

Despite the clear advantages of the proposed TE method for IUP estimation, this can possibly be further improved, e.g. using a physiological model or multiband filtering prior to TE calculation in order to increase estimation accuracy. Additionally, an improvement in IUP estimation accuracy can possibly be achieved by compensating for inter-patient differences of the EHG. This will be the object of future research, taking into account the effects on overall complexity. Finally, an extended dataset should be generated to further validate the obtained results.

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