Uterine EMG spectral analysis and relationship to mechanical activity in pregnant monkeys

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Abstract—*The objective is to analyse internal and external recordings of uterine EMG in order to reveal common features and to assess the relationship between electrical activity and intra-uterine pressure modification. Three monkeys participated in the study, one as a reference and the others for data. EMGs are recorded simultaneously, internally by unipolar wire electrodes and externally by bipolar Ag/AgCl electrodes. Intra-uterine pressure is recorded as a mechanical index. Except for delay measurements, parameters are derived from spectral analysis and relationships between recordings are assessed by studying the coherence. Spectral analysis exhibits two basic activities in the analysed frequency band, and frequency limits are defined as relevant parameters for electrical activity description. Parameter values do not depend on the internal electrode location. Internal and external EMGs present a similar spectral shape, despite differences in electrode configuration and tissue filtering. It is deduced that external uterine EMG is a good image of the genuine uterine electrical activity. To some extent, it can be related to an average cellular electrical activity.*

Keywords—*Electrohysterography, External versus internal recordings, Spectral analysis, Coherence function*

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1 Introduction

ELECTRICAL ACTIVITY represents the trigger of muscle fibre contraction. The resultant mechanical effect depends on the excitation characteristics and the spreading of electrical activity. Uterine smooth muscle cells exhibit negative resting potentials with small and slow spontaneous fluctuations. Isolated action potential or bursts are induced when resting potential fluctuations reach a threshold.

Various methods have been used to record and analyse the electrical activity of the uterus. Both gestation and labour periods have been investigated in mammals and humans. Electromyographic (EMG) signals have been recorded either internally or externally with electrodes located on the abdomen, providing the electrohysterogram (EHG). Many authors have attempted to characterise EMG and EHG (BODE, 1931; DILL and MAIDEN, 1946). Larks *et al.* characterised a female EHG by its shape (LARKS *et al.*, 1957). Their signals were biphasic, starting with a negative deflection, because they were filtered between DC and 1 Hz. However, most authors have based their studies on higher frequency bands (fast wave, above 0·2 Hz) observed in both EMG and EHG, considering slow potential deflections (slow wave) as artefacts (STEER and HERTSCH, 1950; SUREAU *et al.*, 1965). Wolfs and van Leuween observed that EMG activities were well synchronised with IUP peaks, and they measured an EMG propagation velocity of around 20 mm s^{-1} (WOLFS and VAN LEUWEEN, 1979). In work on monkeys, it has been observed that the frequency content of an EMG burst varied within the burst, and also depended on individuals and on recording time (GERMAIN *et al.*, 1982). Planes *et al.* aimed to characterise the EHG by autoregressive modelling to study progress of labour in humans (PLANES *et al.*, 1984). Marque *et al.* demonstrated that there were two frequency bands (FWL (fast wave low) and FWH (fast wave high) within 0·2–3 Hz range (identified as the fast wave) for the human EHG (MARQUE *et al.*, 1986). Inefficient gestational contractions exhibited frequencies mainly located in the lower band, whereas labour contractions had a relative higher frequency content in the higher band. In most studies, electrical activity was used as a marker to indicate the presence or absence of uterine contractile activity. However, there have been few in-depth studies of the true characteristics of this signal.

This work focuses on the description of the two components (FWL and FWH) of the fast wave (DEVEDEUX *et al.*, 1993). Its first objective is to analyse the EMG by describing spectral parameters of the signal recorded on different sites of the

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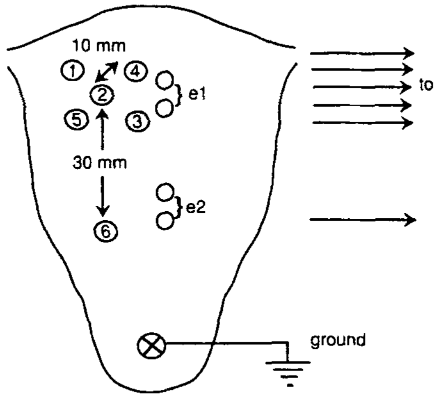

Fig. 1 *Electrode locations on uterus wall (1–6) and externally (e1, e2)*

uterine muscle during pregnancy, by means of internal and external electrodes. The second objective is the analysis of the qualitative relationship between the electrical and mechanical activity of the uterus. The third objective is the investigation of the mode of propagation of the EMG through the uterine muscle.

For clarity, internal (EMG) and abdominal (EHG) electrical activities are referred to as internal and external EMG, respectively. In addition, the generic term of mechanical activity refers exclusively to an increase in the intra-uterine pressure. The two parts of the analysed wave are referred to as FWL and FWH.

2 Material and methods

2.1 *Electrodes*

One cynomolgus monkey (Macaca fascicularis) in the last third of gestation (140–150 days) is studied as a reference to achieve the signal characterisation. Contractions of two other animals in the same gestational state are used as supplementary signals. These two animals participate in another study concerning drug effects on the uterine contractility. However, at the beginning of the experiment, they exhibit some spontaneous contractions which are used to produce additional signals. The animals are housed at 22°C, with 14 h of light and 10 h of darkness. They receive a standard pellet diet twice daily with a fruit supply and tap water *ad libitum*. The general husbandry conditions meet the prescribed European conditions for laboratory animals.

In order to record internal electromyograms, six unipolar electrodes are attached to the surface of the uterine corpus, according to a previously published method (GERMAIN *et al.*, 1982). Internal electrodes are made of double-stranded insulated stainless wires (nickel–chrome blending 80/20), with a cross-section of 120 μm. The distal tips of the wires are stripped for a distance of 2 mm. The wires provide a very acceptable low-noise signal by inserting one strand of each electrode into its correct site on the uterine mûscle and attaching it tightly by nylon threads. The other wire is inserted into the bottom of the cervix and connected to the electrical ground (Fig. 1).

External EMG is recorded differentially by two Ag/AgCl surface electrodes, placed on the abdomen with an inter-electrode distance of 15 mm after the skin has been carefully cleaned. This differential configuration is imposed by a poor signal-to-noise ratio (SNR) in unipolar conditions. Electrode pairs are placed 38 mm apart on the median vertical axis of the uterus, right over the internal electrodes. External signals are recorded at two sites to estimate a possible delay in the uterine electrical wave propagation, with a reference electrode on the flank.

Intra-uterine pressure (IUP) recordings are made by inserting an open-ended polyethylene tube, filled with physiological saline solution, transmyometrically into the amniotic cavity, away from the implantation sites of the placenta*. The tube is then secured to the uterus by a suture. Both internal electrodes and catheter are exteriorised through the flank. The electrodes are directly connected to the acquisition module, whereas the catheter is connected to the same module via a pressure transducer. During recordings, the transducer is maintained at a level midway between the symphysis pubis and the fundal edge of the uterus.

Just prior to full recovery, the animals are placed in a restraining chair, in which they remained for the duration of the study. The risk of premature labour due to the operational procedure is prevented by intramuscular administration of the non-steroidal anti-inflammatory compound Dichlofenac† during the first 24 h (3 mg kg h^{-1}).

2.2 *Acquisition module*

Internal EMG signals are directed to a commercial DC amplifier (EEG polygraph, ECEM) with high input impedance (100 MΩ). External bipolar electrodes are connected through a home-made differential isolated amplifier using an AD589J circuit§, with 100 MΩ input impedance and 120 dB common mode rejection ratio. All EMGs are band-pass filtered (0·2–10 Hg, eighth-order Butterworth). Signals are then digitised at a sampling rate of 30 Hz.

2.3 *Signal description*

Signal recordings are obtained during spontaneous contractions two days after surgery, during a five day period, between 10 a.m. and 5 p.m. The contractions to be processed are selected as those for which both internal and external EMGs, as well as IUP signals, are simultaneously available. Given the complexity of such experiments and regarding the objectives of the study, signal characterisation is achieved on signals from only one animal. Results are then confirmed with signals from the two other animals when it is possible to record spontaneous contractions.

From the animal used as a reference, we obtain simultaneously the relevant EMG and IUP signals for 12 contractions. However, in this series of recordings the signals from electrode 5 present a very poor SNR, probably for bad connection reasons. Hence they are discarded. The five remaining internal signals are divided into two spatially distinct groups: group 1 contains the set of 48 signals derived from electrodes 1, 2, 3 and 4 (near locations); group 2 forms the set of 12 signals derived from electrode 6 located about 30 mm from the mean position of the electrode set of group 1. It is assumed that this distance is sufficiently large to consider the signals detected by these two groups to be distinct (YOSHIHITO *et al.*, 1990).

2.4 *Power spectral density*

Most EMG studies have analysed temporal or spectral features of the signal (DUCHÊNE and GOUBEL, 1993). Among the various signal characterisation methods, the power spectral density (PSD) represents a powerful tool, commonly used for the estimation and analysis of spectral components of temporal

* Statham pressure transducer
† trademark
§ Analog Devices

processes. There exist many ways of estimating this spectral function (KAY and MARPLE, 1981). In our study, the one-sided PSD has been estimated using the Welch method (SHIAVI, 1991; WELCH, 1970), in which the squared modules of the Fourier transforms $X_k(f)$ of successive 50% overlapped EMG epochs are averaged

$$P_{xx[fn]} = \frac{2}{KN\Delta t}\sum_{i=1}^{K}|X_i(fn)|^2, \qquad n = 0, 1, \ldots, N/2$$

where Δt is the sampling period, N is the total number of samples and K is the number of epochs used in averaging (BENDAT and PIERSOL, 1986).

PSD estimation can only be achieved in stationary conditions. Therefore, signals are preprocessed for stationarity assessment using reverse arrangement and autocorrelation tests (BENDAT and PIERSOL, 1986). Stationary time intervals are found to have a mean length of 501 samples ($N = 19$, s.d. = 107). PSD is then estimated over the longest stationary intervals included within each burst of electrical activity using three overlapped Hamming-windowed epochs of 256 samples each, approaching Welch's condition.

2.5 *Coherence function*

The coherence function is generally used to quantify the alterations that occur in the strength of association and timing between two processes brought about by the presence of other factors (CARTER, 1987; ROSENBERG *et al.*, 1989). This function is expressed as

$$C(f) = \sqrt{\frac{|P_{xy}(f)|^2}{P_{xx}(f)\cdot P_{yy}(f)}}$$

where $P_{xx}(f)$ and $P_{yy}(f)$ are the expectation (statistical mean) of the power spectra of each signal, and $P_{xy}(f)$ is the expectation of their cross-power spectra. $C(f)$ varies in the [0, 1] interval, reflecting the amount of linear association of the two signals at each frequency value of the spectrum.

The coherence is estimated between internal electrodes of the same group (electrode 2 versus electrode 3), between electrodes of different groups (electrode 2 versus electrode 6) and between internal and external electrodes (electrode 2 versus electrode e1). As for PSD estimation, signals are first preprocessed for stationarity assessment. Each spectral density is then estimated from three overlapped windows to produce 256 spectral points (the overlapping rate depending on the stationarity interval length). Finally, expectations are computed from the signals of the 12 reference contractions.

2.6 *Spectral parameters description*

In order to characterise properly the EMG spectral shape on the basis of the two waves FWL and FWH, spectral parameters are chosen:

(*a*) SF, position of the minimum between FWL and FWH.
(*b*) MAX1, limit for which the band (0·2–MAX1) includes 95% of FWL power.
(*c*) MIN2 and MAX2, symmetrical limits including 95% of FWH power (2·5% bilateral tail areas).
(*d*) MED1 and MED2, median frequencies of FWL and FWH, respectively.
(*e*) MPF1 and MPF2, mean frequencies of FWL and FWH, respectively.
(*f*) NRJ1 and NRJ2, relative power of FWL and FWH, respectively.

These parameters are computed for each contraction and each electrode.

2.7 *Propagation analysis*

An important point to assess is the relationship between electrical activity recorded internally and externally on separated sites on the uterus, and the mechanical activity, on the basis of the difference between their onset times. This needs to be done to investigate the overall propagation delay through the uterine muscle, and to study the features of the temporal relationship between electrical and mechanical activities.

The same 12 contractions are selected for this study. However, the recording of related mechanical activity is too noisy for one contraction (poor accuracy on the onset time), leaving 11 usable contractions. The analysis is applied separately on slow and fast components of internal recordings. For external recordings, only the fast component is retained, as the slow one is affected by many contaminating noises such as mechanical artefacts.

Slow and fast activities are separated by band-pass digital filtering. In order to determine the EMG onset time, quasi-instantaneous signal power is estimated by time record squaring and slight smoothing (seven point moving average filter). Under these conditions, the onset time is determined as the instant at which this power estimate reaches 5% of its maximum value. Thus, 44 values of onset time are evaluated for the signals of group 1, and 11 values for group 2, for both slow and fast activities. For external signals, 22 values of onset time are evaluated, corresponding to the two external electrode sets.

A linear regression is applied on a manually selected window centred on the rising slope of the IUP curve. This allows the definition of the onset time of pressure as the intersection point between the resting pressure level and the resulting model. Taking the onset time of pressure as a time reference, the delay between electrical and mechanical activities is computed for each signal.

2.8 *Statistics*

Parameters extracted from the two groups of internal signals are statistically compared on the basis of a test of difference in sample means. The student t-test (unpaired case) is used with a significance level $p = 0{\cdot}05$. The same test is then applied to compare parameter values extracted from internal and external recordings.

The delay values between two different electrodes are compared by the Wilcoxon rank test with the same significance level $p = 0{\cdot}05$.

3 Result

3.1 *EMG characterisation*

Fig. 2*a* shows a recording of uterine electrical activity derived from an internal electrode (upper tracing), an external electrode (middle tracing) and the related IUP curve. In almost all detected contractions (selected or not), an increase in internal pressure is associated with the presence of an electrical signal in both internal and external recordings. In less than 1% of cases, the presence of electrical activity without recorded mechanical effect is noticed.

The average of the 48 PSDs of Group 1 signals, computed as indicated above, is shown in Fig. 3 (semi-log scale), where the frequency range has been limited to 0·2–10 Hz. FWL represents the activity between 0·2 and 1·2 Hz. FWH peaks at around 3 Hz.

In regard to temporal characteristics, FWL appears to be synchronised with the onset of the mechanical activity and

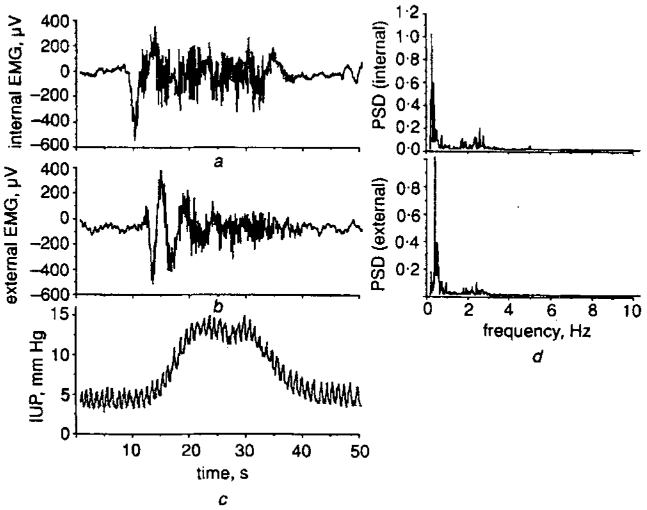

Fig. 2 *Spontaneous electrical activity of monkey uterus, at late gestation; (a–c) raw signals recorded by unipolar internal electrode (a) and bipolar external electrode (b), with their associated IUP curve (c); (d) corresponding power spectral densities (arbitrary units)*

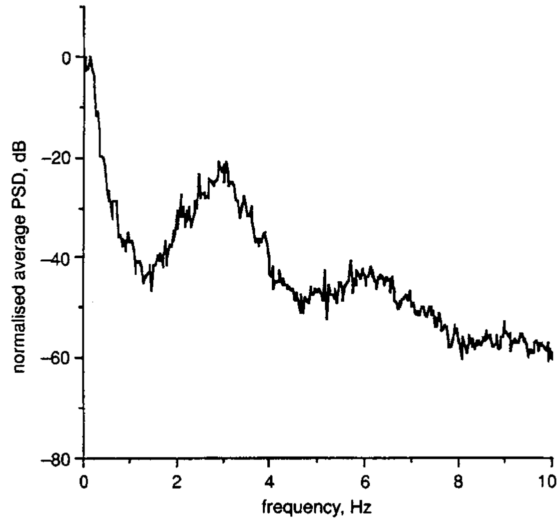

Fig. 3 *Average of 48 PSDs of EMGs; note the presence of FWL (activity under 1·2 Hz) and FWH (activity around 3 Hz); periodic aspect of the spectrum is enhanced by this semi-log representation*

remains present through the whole contraction duration. Meanwhile, FWH is restricted to the rising edge of the IUP curve and lasts a few seconds after the peak of the contraction.

When exploring the spectral features in detail, a kind of periodicity at high frequencies (Fig. 3) can be noticed, which results in a repetition of the spectral part identified as FWH. This type of spectral shape can arise when a non-sine carrier is frequency-modulated. Hence, this phenomenon is assumed to be due to changes (modulation) in the firing rate of cellular potential spikes (KAO, 1977).

Based on this assumption of frequency modulation, the relevant part of FWH can be limited to the first peak (fundamental), with the boundary computed as the minimum position between the fundamental and the second peak (first harmonic). This boundary would have been a good parameter for spectral estimation. Unfortunately, if harmonics can often be shown on internal recordings, they seldom appear on external recordings, making it impossible to define any boundary. Therefore, we decide to compute a fixed limit as the minimum between fundamental and first harmonic from the mean spectrum of all available internal signals, where it is possible to determine such a limit. This leads to a frequency value of 4·7 Hz. From this point, it becomes possible to compute the spectral parameter values as defined in Section 2.6 (Table 1).

These results indicate that there is no difference between the various parameters of EMG due to electrode location, except for the position of the limits of FWH. In addition, no significant difference remains for $p < 0{\cdot}01$. Thus, we are justified in pooling the signals from all internal electrodes, yielding the values given in the last row of Table 1.

In order to compare the spectral parameters of internal and external recordings with as many external signals as possible, the foregoing reference set is enlarged by adding signals from six supplementary contractions where only two internal signals are recorded, then unadapted to the previous analysis. Table 2 summarises the results of this comparison. On the basis of these, although internal and external activities are found to have the same morphologic aspects in their spectra, differences are noted in the parameters of FWL, as well as in the relative power of the two main activities. Fig. 4 shows the coherence functions involving various internal and external signals. A higher coherence level can be noticed between internal electrodes around the main peaks of the fast activity. However, there is no coherence between internal and external recordings.

3.2 *Relationship between electrical and mechanical activities*

The delays are computed for the 11 reference contractions, where five internal and two external recordings are available simultaneously with the IUP. For the five internal electrodes, FWL and FWH were analysed separately (Figs. 5*a* and *b*, respectively) in order to test if they have specific roles with respect to the propagation process. On the other hand, the whole signal is analysed for the two external electrodes (Fig. 5*c*). The Wilcoxon rank test is applied to the delays computed from electrodes 1 and 6 (the most distant internal electrodes), revealing no significant difference in delay.

The fact that time delay values are all positive validates the methodology, confirming that electrical activity originates earlier than amniotic pressure onset at every location in the uterus. In addition, the study of time delays between one

Table 1 Spectral parameters of internal unipolar recordings

parameter	MED1	MPF1	MAX1	NRJ1	SF	MIN2	MED2	MPF2	MAX2	NRJ2
Group 1	0·32 ± 0·08	0·37 ± 0·12	1·26 ± 0·4	0·43 ± 0·2	1·50 ± 0·29	1·84 ± 0·24	3·11 ± 0·23	3·01 ± 0·18	4·24 ± 0·18	0·57 ± 0·17
Group 2	0·35 ± 0·11	0·39 ± 0·14	1·25 ± 0·38	0·44 ± 0·2	1·52 ± 0·24	1·92 ± 0·27	3·1 ± 0·27	3·03 ± 0·2	4·35 ± 0·08	0·56 ± 0·2
test result	=	=	=	=	=	=	=	=	≠	=
mean value	0·32 ± 0·09	0·37 ± 0·12	1·26 ± 0·39	0·43 ± 0·03	1·50 ± 0·3	1·86 ± 0·25	3·11 ± 0·24	3·01 ± 0·19	4·26 ± 0·18	0·57 ± 0·03

values are indicated as mean ± SD (in Hz for frequency limits) for groups 1 and 2, and results of the comparison are based on student t-test ($p = 0{\cdot}05$); $N = 48$ for group 1 and $N = 12$ for group 2; NRJ1 and NRJ2 are relative to total power

Table 2 Spectral parameters of internal and external recordings

parameter	MED1	MPF1	MAX1	NRJ1	SF	MIN2	MED2	MPF2	MAX2	NRJ2
internal	$0{\cdot}35 \pm 0{\cdot}13$	$0{\cdot}40 \pm 0{\cdot}15$	$1{\cdot}28 \pm 0{\cdot}39$	$0{\cdot}45 \pm 0{\cdot}03$	$1{\cdot}50 \pm 0{\cdot}29$	$1{\cdot}84 \pm 0{\cdot}27$	$3{\cdot}02 \pm 0{\cdot}3$	$2{\cdot}95 \pm 0{\cdot}24$	$4{\cdot}27 \pm 0{\cdot}18$	$0{\cdot}56 \pm 0{\cdot}03$
external	$0{\cdot}40 \pm 0{\cdot}07$	$0{\cdot}45 \pm 0{\cdot}08$	$1{\cdot}36 \pm 0{\cdot}34$	$0{\cdot}72 \pm 0{\cdot}02$	$0{\cdot}72 \pm 0{\cdot}33$	$1{\cdot}99 \pm 0{\cdot}36$	$2{\cdot}94 \pm 0{\cdot}53$	$2{\cdot}92 \pm 0{\cdot}44$	$4{\cdot}25 \pm 0{\cdot}25$	$0{\cdot}29 \pm 0{\cdot}02$
test result	$\neq$	$=$	$=$	$\neq$	$\neq$	$\neq$	$=$	$=$	$=$	$\neq$

$N = 72$ for internal and $N = 36$ for external (same test conditions as Table 1).

location and another does not provide any specific trend in the propagation of the electrical activity on the whole uterine muscle. In these experimental conditions of spontaneous contractions in late pregnancy, it is thus impossible to identify a unique well defined source of activity that propagates along the muscle.

In regard to surface electrodes, the observed time delay values measured between IUP onset and the recorded activities for the two external electrodes show that all electrical activities precede the corresponding mechanical activity, except for the first contraction. In addition, a close similarity can be noticed between shapes of curves obtained from internal and external electrical activities. Even though the small number of points precludes computing a significant correlation, there is nevertheless a close similarity between internal and external recordings.

3.3 *Additional results*

Additional signals are processed from the two other monkeys when it is possible to select spontaneous contractions. Ten and nine contractions are selected from supplementary monkeys 1 and 2, respectively. Concerning supplementary monkey 1 (Supp. 1), electrode polarisation made it necessary to change the high-pass cut-off frequency to a higher value during the experiment. Therefore, comparison of FWL activities becomes impossible, limiting parameter assessment to FWH. Parameter values are summarised in Table 3, as well as the result of comparison with the reference, based on the student's t-test ($p = 0{\cdot}05$).

4 Discussion

Most studies on uterine contractile activity have handled the EMG signal in a qualitative manner, with its description often restricted to the presence or absence of electrical activity, related to IUP. This study is an effort to characterise and quantify the internal unipolar signal recorded from different sites of the uterine muscle on the basis of its spectral parameters. It also identifies the kind of relationship between internal and external recordings using the IUP as a reference.

In terms of EMG characterisation, the frequency values derived in our study are sometimes slightly higher than those described by other authors (MARQUE *et al.*, 1986; PLANES *et al.*, 1984), especially with respect to FWH. These differences from the values given by other authors, who have explored external recordings of electrical activity of the human uterus, can be attributed to the difference in size between the monkey uterus and the human uterus, which classically leads to higher frequency values for smaller muscles (STAHL, 1967). On the other hand, the difference found on MAX2 for the two groups of internal electrodes (Table 1) may be due to the sensitivity of this parameter to contaminating noise (poor signal-to-noise ratio in this frequency area).

Results from additional signals indicated in Table 3 first confirm that parameters derived from power or energy computation are not relevant, owing to their variance with respect to many environmental conditions like electrode configuration, position inside the muscle or tissue filtering. On the other hand, results on characteristic frequencies present a good agreement between reference and Suppl. 2, especially for FWH which seems to be the more interesting information as an image of average cellular electrical activity. They are more disappointing for Suppl. 1, except for the lower limit of FWH. However, the experimental conditions make it difficult to draw a conclusion in that case (as indicated before, large baseline fluctuations made it necessary to change the filtering characteristics during the experiment). However, the results are presented here because the signals exhibit the same qualitative characteristics (well separated slow and fast activities, harmonic components of FWH).

The comparison of the spectra obtained from internal and external recordings reveals a qualitative similarity between the spectral shapes. However, it is difficult to achieve this comparison in a quantitative manner, owing to the differences

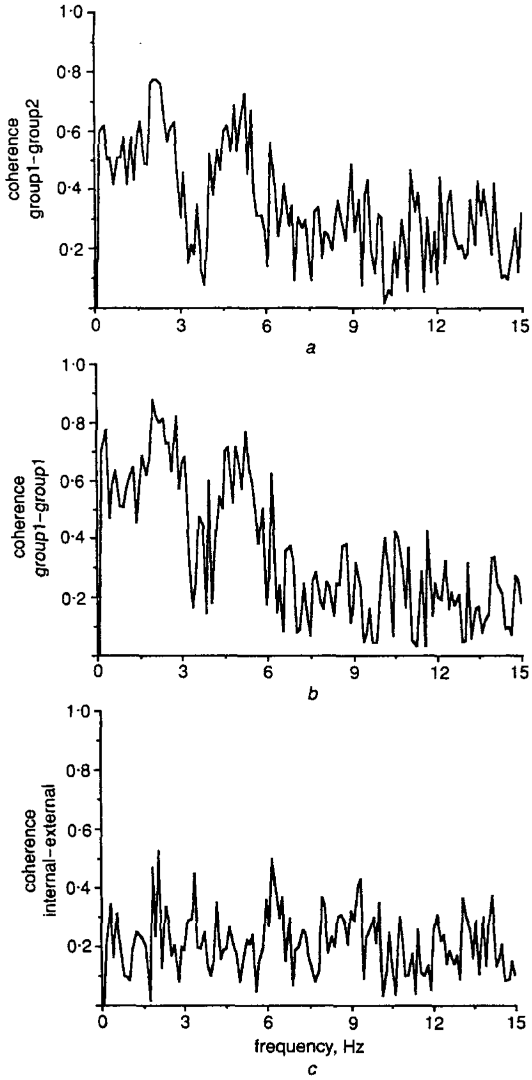

Fig. 4 *Coherence functions computed between internal signals derived from different groups (a), from the same group (b), and between internal and external recordings (c)*

Table 3 Comparison with additional internal recordings

parameter	MED1	MPF1	MAX1	NRJ1	SF	MIN2	MED2	MPF2	MAX2	NRJ2
reference	$0{\cdot}35 \pm 0{\cdot}13$	$0{\cdot}40 \pm 0{\cdot}15$	$1{\cdot}28 \pm 0{\cdot}39$	$0{\cdot}45 \pm 0{\cdot}03$	$1{\cdot}50 \pm 0{\cdot}29$	$1{\cdot}84 \pm 0{\cdot}27$	$3{\cdot}02 \pm 0{\cdot}3$	$2{\cdot}95 \pm 0{\cdot}24$	$4{\cdot}27 \pm 0{\cdot}18$	$0{\cdot}56 \pm 0{\cdot}03$
Suppl. 1					$1{\cdot}34 \pm 0{\cdot}41$	$1{\cdot}79 \pm 0{\cdot}37$	$3{\cdot}38 \pm 0{\cdot}53$	$3{\cdot}18 \pm 0{\cdot}31$	$4{\cdot}78 \pm 0{\cdot}10$	
test result					$=$	$=$	$\neq$	$\neq$		
Suppl. 2	$0{\cdot}45 \pm 0{\cdot}09$	$0{\cdot}45 \pm 0{\cdot}10$	$1{\cdot}45 \pm 0{\cdot}34$	$0{\cdot}81 \pm 0{\cdot}08$	$1{\cdot}69 \pm 0{\cdot}25$	$1{\cdot}88 \pm 0{\cdot}26$	$2{\cdot}97 \pm 0{\cdot}44$	$2{\cdot}98 \pm 0{\cdot}32$	$4{\cdot}8 \pm 0{\cdot}17$	$0{\cdot}19 \pm 0{\cdot}08$
test result	$\neq$	$=$	$=$	$\neq$	$=$	$=$	$=$	$=$	$\neq$	$\neq$

$N = 10$ for Supp. 1 and $N = 9$ for Suppl. 2; for Suppl. 1, electrode polarisation made it necessary to change the high-pass cut-off frequency during the experiment, limiting the parameter assessment to FWH (same test conditions as Table 1)

in the experimental recording conditions. It is necessary to record the internal signals in a unipolar manner, in order to be able to conclude about the invariability of the spectral parameters with respect to internal electrode locations. As a matter of fact, it is unthinkable to reproduce the same differential configuration in many locations using wire electrodes. On the other hand, unipolar external recordings produce a very poor SNR and consequently unexploitable data, leading to the use of a differential configuration. This difference in the recording method associated with obviously distinct transfer functions between signal source and internal or external electrodes can easily explain some discrepancies in the spectral shapes.

The size of the external electrodes has an integrating effect on the recorded signal, thus a low-pass filtering effect and an increase in the overall recorded energy (HELAL and BOUISSOU, 1993). The distance between the signal source and the recording location also acts as a low-pass filter, but produces a concomitant signal attenuation. Filtering due to a propagation phenomenon could be considered only on the external differential electrode set. When electrical events are propagating at a constant velocity under the electrodes, the equivalent filter is high-pass. In our case, no propagation has been proved internally. Therefore, this specific effect of a differential electrode configuration only applies to signals propagated on

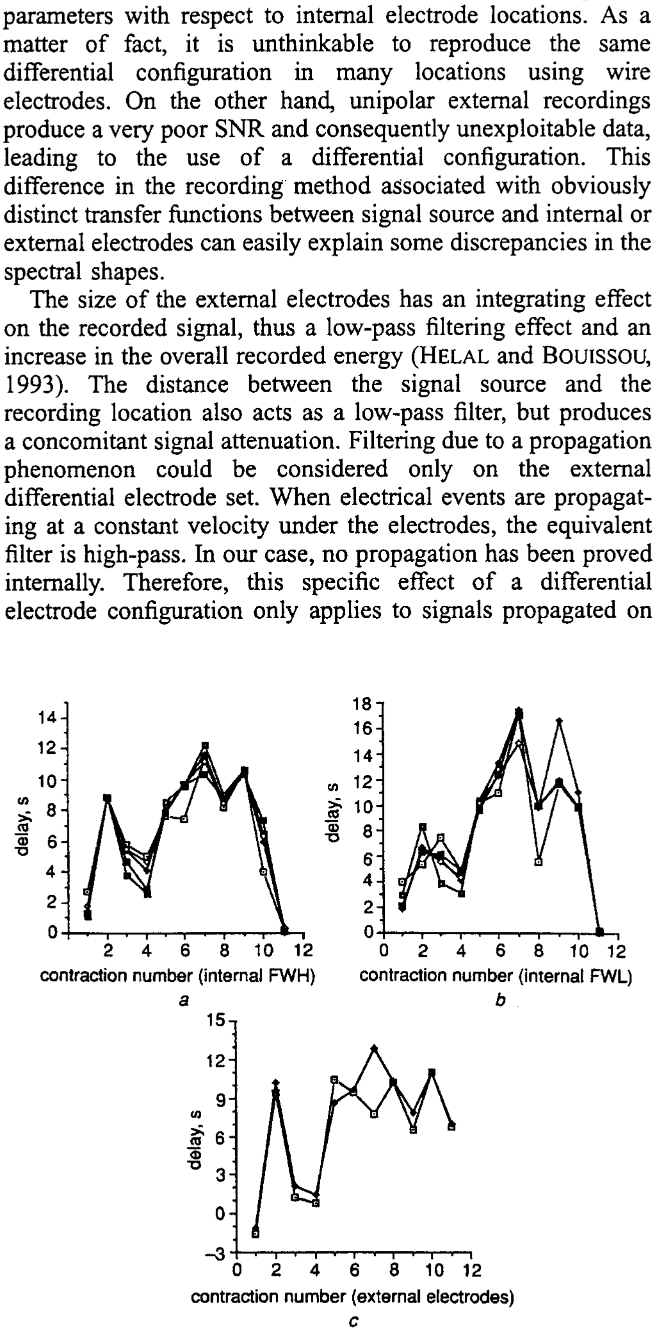

Fig. 5 *Time delay values between electrical activity arrival times under different electrodes and mechanical activity onset, for (a) internal FWH, (b) internal FWL and (c) external recordings; note that the delays are all positive except for one external recording, showing that electrical activity always precedes mechanical activity*

the abdominal surface, thus to low-frequency propagated artefacts. These points mainly explain the differences between internal and external signals in their spectral power distribution (NRJ1 and NRJ2). The difference between the SF values can be supposed to be the consequence of the lower signal-to-noise ratio of FWH of external signals, leading to a larger variance in the computation of this parameter. This variance obviously influences the MIN2 parameter computation to the same extent.

However, it is clear that the signals have the same overall characteristics, external or internal, whenever they are recorded on the uterine surface. In addition, it is worth noting that the coherence function exhibits a significant increase for internal signals in the bands located around the two main parts of FWH. On the other hand, the weakness of the coherence function level between internal and external signals reflects the complexity and the nonlinearity of the corresponding propagation function. Nonetheless, the results on the shapes of the PSDs and the good correspondence between characteristic spectral parameters indicate a close similarity between internal and external electrical activities, and the validity of studies on the contractile process and the uterine electrical activity by means of external recordings (MARQUE *et al.*, 1986).

Our results (Fig. 5) on the positive delay between mechanical and electrical activities show the precedence of the EMG signal wherever it is recorded, with respect to the IUP increase, i.e. the mechanical process. This confirms qualitatively the temporal correlation between both activities, and indicates that the pressure increase is directly related to the recorded electrical activity. The temporal correspondence between IUP and both external and internal EMGs reinforces the evidence of this relationship.

Considering the complexity of the EMG signal and the nonlinearity of the propagation function, the classical and linear tools usually used for the exploration of propagation phenomena through the myometrial muscle (e.g. the cross-correlation function) can no longer be correctly applied. The only possible way to study the propagation mode implies the use of methods based on the EMG 'group delay' measurement, with IUP onset taken as the time reference. On the other hand, it is impossible to determine a well defined group propagation direction. Complexity and probable immaturity of the propagation medium during this period of gestation might explain the absence of privileged propagation direction. In many species, the cell-to-cell communication via gap-junctions is established only when parturition is imminent (GARFIELD, 1986).

Results on the invariability of spectral parameters of internal EMG signals recorded from different sites favour the hypothesis of spatial unity of the characteristics of this signal; a conclusion which suppresses any restraint on the position of the recording electrodes, internally on the uterine corpus (excluding the cervix area), in order to assess uterine electrical activity. This equivalence of the parameter values all along the muscle provides a unique EMG model, when recorded during late pregnancy, under spontaneous contractions.

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