RECORDING OF UTERINE ACTIVITY FROM THE ABDOMINAL LEAD EMG

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INTRODUCTION

Monitoring uterine activity is important in prenatal medicine for surveying the course of the pregnancy and evaluating the condition of the fetus. Two methods, one internal and the other external, are known for recording uterine activity. With both techniques the activity is evaluated in terms of the uterine pressure, the tocogram.

For internal tocography, the pressure is determined by introducing a fluid-filled open-end catheter into the uterine cavity. It is possible with this method to determine the absolute intrauterine pressure. Often however, internal tocography is rejected because of the practical difficulties involved. Moreover, very recent tests indicate that the use of a catheter may influence parturition, and so it should not be used in premature deliveries where monitoring of the contraction activity would be of particular significance.

With external tocography some form of pressure or force transducer is fastened to the abdominal wall by means of an elastic belt. Uterine contractions produce changes in the displacement of a sensor pin which acts on the transducer, producing an electrical signal that corresponds to the relative uterine pressure. External tocography, however, has the drawback that it is subject to many factors which may influence the results of the measurement. We feel that the limitations of tocography cannot be eliminated by technical improvements of the measuring equipment, and therefore the possibility of recording the uterine contractions from the abdominal lead electromyogram (EMG) without losing any information regarding tocography has been studied. As a result of the study it has become evident that the recording of uterine activity using the EMG is more reliable than tocography and offers even additional information about the excitation and propagation of the contractions. The basic methods that have been developed and the results of clinical tests are presented here.

THE RELATION BETWEEN SURFACE EMG AND INTRAUTERINE PRESSURE

Changes in intrauterine pressure are produced by contractions of the uterine muscle. Any contraction is stimulated by a series of action potentials. The force that is developed by the muscle depends on the number of activated, i.e. depolarising, fibres within the muscle and on the frequency of stimulation of each individual fibre. The time constant of the contraction is large compared to that of the stimulating pulse, and so the resulting force can be written as:

$$f(t) = c_1 \cdot \sum_{n=1}^{N} s_n(t)$$

where c_1 is a constant, N is the total number of fibres in the muscle and $s_n(t)$ is the firing rate of the n-th fibre.

The linear summation is not really valid because of the non-linear properties of the muscle fibres. However, this is a good first approximation and is sufficient to describe the basic relations. The change of the intrauterine pressure that results from the force of the uterine muscle during contractions can be approximated using the simple model of a fluid-filled balloon, where pressure is propagating equally in all directions and muscular force is directed tangentially to the surface of the balloon. It is well known that the change of the internal pressure of the balloon is proportional to the external force, i.e.

$$\Delta p(t) = c_2 \cdot f(t) \tag{1}$$

where c_2 is a constant. The absolute intrauterine pressure can be written as:

$$p(t) = p_0 + c_2\, f(t) = p_0 + c_3 \cdot \sum_{n=1}^{N} s_n(t) \tag{2}$$

where p_0 represents the base-line pressure.

When a muscle fibre depolarises an action potential is produced which travels along the fibre. The shape of the action potential when recorded depends on fibre size, the propagation velocity, and the geometric relationship of the measuring electrodes to the fibre. By applying the electromagnetic field theory, the electrode potential can be calculated for any given geometry.

The biphasic action potentials of the individual muscle fibres are summated at a surface electrode, and the recorded potential is called the electromyogram (EMG). As the stimulation of the different fibres is not synchronised, a complicated interference pattern is observed which is very sensitive to the firing rate and firing pattern of the individual fibres and to the position of the electrode.

Rectifying and low pass filtering of the EMG produces a measure of its intensity (IEMG). The IEMG measured from a single electrode can be estimated as:

$$\mathrm{IEMG}(t) = c_4 \sum_{n=1}^{N} 1/r_n^2 \cdot \cos\Phi_n \cdot s_n(t) \tag{3}$$

where c_4 is a constant, s_n again denotes the firing rate of the n-th fibre, N is the total number of fibres, r_n is the distance between the n-th fibre and the electrode and Φ_n is the angle between the n-th fibre and the connecting line to the electrode.

This equation for the IEMG assumes that the muscle fibre has a straight cylindrical shape, that its length is small compared with the distance to the electrode and that the conductivity of the body is homogenous. Without this assumption the computation becomes more difficult, but the result would be essentially the same. Another simiplification is that the superposition of the biphasic action potentials of the individual muscle fibres results in a linear dependency of the intensity of the EMG on the number of activated fibres, if all action potentials have equal amplitudes. This is not really valid for all muscles, as the superposition is influenced by the probability distribution for the pattern of stimulation of the single fibres. In the special case of the uterine muscle this assumption is in accordance with the results of experimental studies.

Comparing equations (2) and (3) it becomes evident that there is a high correlation between the intrauterine pressure and the IEMG. The major difference is in the dependency of the IEMG on the electrode position.

If the distance r_n and the angle Φ_n were equal for all fibres the intrauterine pressure changes would be proportional to the IEMG. Unfortunately this condition cannot be fulfilled because of the great extension of the uterine muscle, but it can be approximated by using several electrodes for the measurement. Each of them will register mainly the activity of the nearest parts of the muscle. The individual potentials are then summated. Provided that suitable electrode positions are chosen, the changes in intrauterine pressure and the IEMG are mutually proportional:

$$\mathrm{IEMG}(t) = c \cdot \Delta p(t) \qquad (4)$$

and:

$$p(t) = p_o + 1/c \cdot \mathrm{IEMG}(t) \qquad (5)$$

Using other electrode configurations, the activity of particular parts of the uterine muscle or the propagation of the contractions can be monitored.

The waveshape of the IEMG is strongly dependent upon the time constant of the lowpass filter. Using a small time constant (< 1sec) results in a very ragged wave form. Greater time constants involve a smoother waveform at the expense fo some loss of transient response.

MEASUREMENTS

The abdominal lead EMG of the uterine muscle is disturbed by the maternal ECG, the fetal ECG and the EMG of the abdominal muscle. The maternal and the fetal ECGs can be eliminated by using either appropriate bandpass filters (150–250Hz) or by using the subtraction algorithm described in [1].

As the contractions of the abdominal muscle also result in intrauterine pressure changes, it is not necessary to eliminate its EMG from the abdominally derived signal.

Figures 1, 2 and 3 show a comparison of the uterine activity curves obtained from an external pressure transducer and from the EMG. The EMG has been measured using two electrodes attached to the isthmus uteri and the fundus uteri.

Figure 1 shows the good correspondence between the externally measured

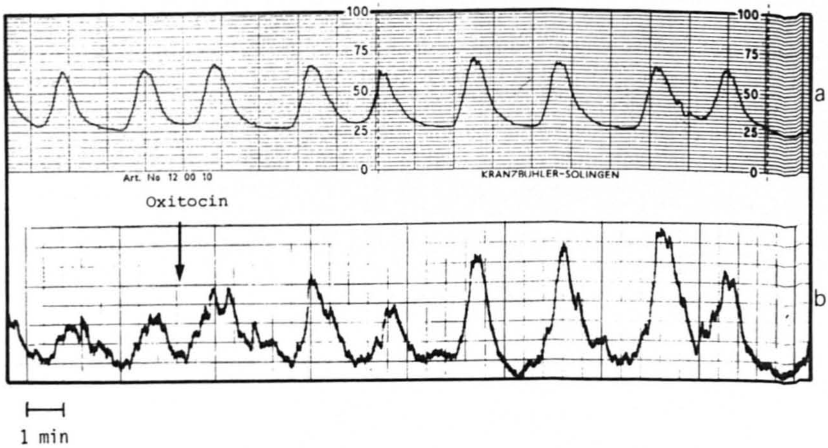

Figure 1. A record of the uterine activity measured by using an external pressure transducer (a) and by processing the EMG (b)

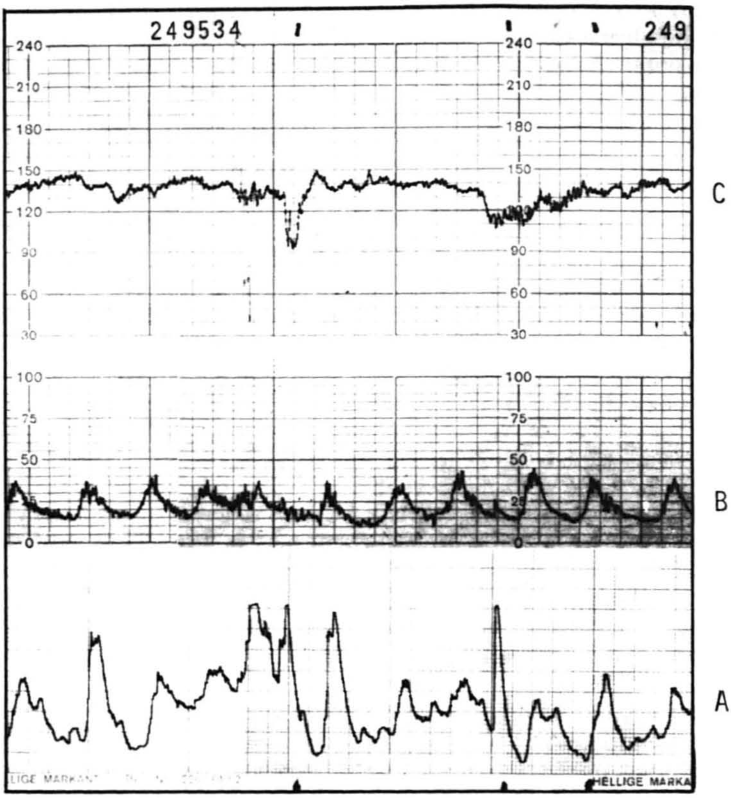

Figure 2. Cardiotocogram. The FHF is computed from the scalp lead ECG (C), the uterine pressure is determined by an external pressure transducer (B) and from the IEMG (A)

pressure and the uterine activity recorded from the IEMG. However, it can be seen that the IEMG contains more information about the progressive increase in the intensity of the contractions as a reaction of the uterine muscle to an oxytocin infusion.

When analysing the cardiotocogram from Figure 2 it is evident that the correlation between FHF and IEMG is greater than between FHF and externally measured pressure.

In Figure 3 the IEMG shows that the excitation of a new uterine contraction starts immediately after the end of the preceding one. This information is lost

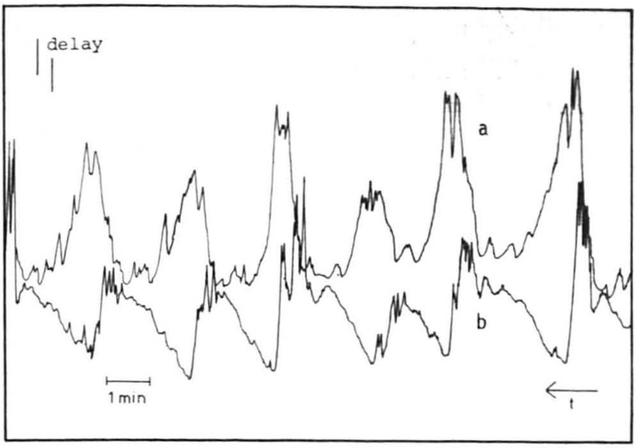

Figure 3. Uterine activity curves measured by using an external pressure transducer (a) and by processing the EMG (b)

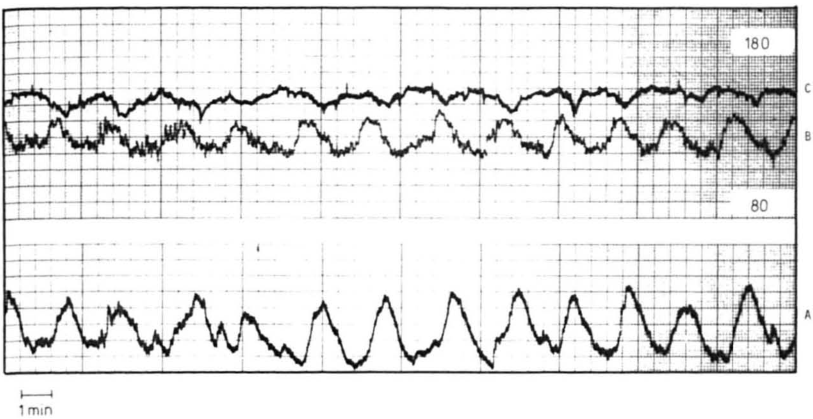

Figure 4. Registration of uterine activity (A), fetal (C) and maternal (B) heart frequency from one single abdominal lead

when using a lowpass filter with a longer time constant.

The particular advantages of recording the uterine activity from the abdominal lead EMG become evident in the case where abdominally attached electrodes are used to obtain signals from the uterus as well as from fetal and maternal cardiac activity (Figure 4). This provides the possibility in an ante-natal examination of obtaining extensive data from a single measurement giving the minimum discomfort to the patient and enabling the physician to obtain an accurate picture of the state of both mother and fetus.

Reference

1 Nagel, J and Schaldach, M (1980) Processing the abdominal fetal ECG using a new method. In *Fetal and Neonatal Physiological Measurements.* (Ed) P Rolfe. Pitman Medical Limited, Tunbridge Wells, UK. Page 9