Ultrasound transducer positioning aid for fetal heart rate monitoring

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Ultrasound Transducer Positioning Aid for Fetal Heart Rate Monitoring

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***Abstract*— Fetal heart rate (fHR) monitoring is usually performed by Doppler ultrasound (US) techniques. For reliable fHR measurements it is required that the fetal heart is located within the US beam. In clinical practice, clinicians palpate the maternal abdomen to identify the fetal presentation and then the US transducer is fixated on the maternal abdomen where the best fHR signal can be obtained. Finding the optimal transducer position is done by listening to the strength of the Doppler audio output and relying on a signal quality indicator of the cardiotocographic (CTG) measurement system. Due to displacement of the US transducer or displacement of the fetal heart out of the US beam, the fHR signal may be lost. Therefore, it is often necessary that the obstetrician repeats the tedious procedure of US transducer positioning to avoid long periods of fHR signal loss. An intuitive US transducer positioning aid would be highly desirable to increase the work flow for the clinical staff. In this paper, the possibility to determine the fetal heart location with respect to the transducer by exploiting the received signal power in the transducer elements is shown. A commercially available US transducer used for fHR monitoring is connected to an US open platform, which allows individual driving of the elements and raw US data acquisition. Based on the power of the received Doppler signals in the transducer elements, the fetal heart location can be estimated. A beating fetal heart setup was designed and realized for validation. The experimental results show the feasibility of estimating the fetal heart location with the proposed method. This can be used to support clinicians in finding the optimal transducer position for fHR monitoring more easily.**

I. INTRODUCTION

Measuring the fetal heart rate (fHR) during labour and delivery is the most commonly applied obstetric procedure to assess the health status of the fetus and to determine whether intervention is required [1]. However, for Doppler ultrasound (US) long periods of signal loss have been reported during fHR monitoring [2]. For the assessment of the fetal condition, these periods limit the interpretation of the heart rate traces in clinical practice. The International Federation of Gynaecology & Obstetrics (FIGO) recommends to accept fHR traces for clinical interpretation only when the total time of signal-loss periods is less than 20% [3].

Since the fetal heart needs to be located within the US beam, Doppler US recording quality highly depends on the correct positioning of the US transducer on the maternal abdomen. In clinical practice, skilled nurses palpate the maternal abdomen and determine the fetal presentation. While relying on a signal-quality indicator and listening to the Doppler audio output of the cardiotocographic (CTG) monitoring system, the US transducer is moved over the maternal abdomen until an acceptable signal is found. Obtaining a good signal is not always possible and real-time imaging techniques are sometimes needed for further assistance in finding the location of the fetal heart [4]. Thus, finding the optimal transducer position is a tedious task that requires advanced skills.

During long fHR monitoring periods it is likely that either the US transducer is displaced or the fetal heart moves out of the US beam. Consequently, the clinical staff needs to reposition the US transducer typically multiple times to minimize the signal loss periods.

A method that supports transducer positioning will help to reduce the occurrence of signal loss. The objective of this work is to investigate the feasibility of using a commercially available CTG US transducer, with only a limited number of transducer elements, to determine the relative location of the fetal heart within the measurement volume of the US transducer. The proposed approach is based on identifying the element, which receives the strongest Doppler signal. Given the array aperture of the transducer, an estimation of the fetal heart location can then be obtained.

II. MATERIAL AND METHODS

A. System Modelling

In a commercially available CTG US transducer (Philips Avalon, Philips Medizin-Systeme Böblingen GmbH, Germany), a limited number of transducer elements are positioned in a circular arrangement. The array aperture A of such a transducer is depicted in Fig. 1. For location estimation of the fetal heart, which is solely based on the received Doppler signal power in the transducer elements, it is required that the received power can unambiguously be linked to a fetal heart location. When the transmitted US wave reaches the fetal heart, the reflected wave travels back towards each transducer element. Ideally, the transmitted US wave is a plane wave, such that the strength of the received signal depends only on the location of the fetal heart and the position of the receiving transducer elements. A radiation pattern with multiple side lobes would make an unambiguous location estimation challenging. Due to the finite size of the array aperture, generation of a plane wave is however not possible.

In this section it is investigated how to optimally drive the array aperture, in order to create a broad radiation pattern without large side lobes. In order to keep the transmission

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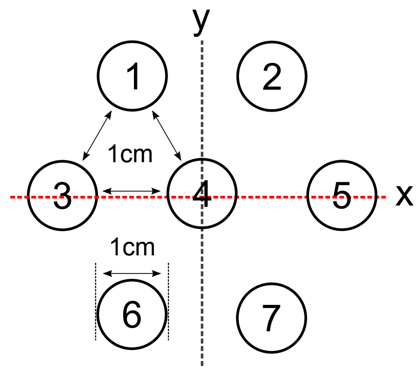

Fig. 1: Array aperture A of the CTG ultrasound transducer. The transducer consists of seven transducer elements positioned in a circular arrangement.

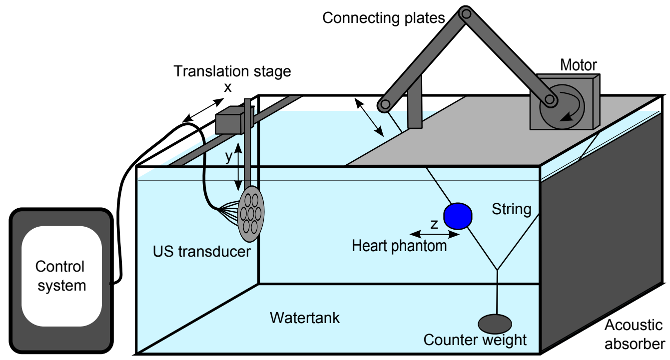

Fig. 2: Schematic drawing of the experimental setup, in which the displacement of a moving fetal heart can be mimicked.

phase simple, beam steering and focussing is not considered at this stage. Two different transmit configurations are explored in detail, viz. transmitting with all elements simultaneously or using only the middle element (element 4 in Fig.1) for transmission of an US wave.

To model the radiation pattern σ of the system, Huygens' principle is used:

$$p_t(X,Y) \propto \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} A(x,y)e^{-j2\pi(xu+yv)}dxdy. \quad (1)$$

According to Eq. (1), the transmitted pressure field p_t in a XY-plane parallel to the array aperture A, see Fig. 1, is proportional to the two dimensional Fourier transform of A, with $u = X/L\lambda$, $v = Y/L\lambda$, L the distance to the plane and λ the wavelength of the transmitted center frequency f_0.

The two investigated approaches differ only in their transmission phase. During the receive phase, in both approaches each transducer element, indicated by the element number n, receives the reflected US waves individually. The receive beam $p_{r,n}$ is identical for all elements n, except that the center of the beam is shifted to the spatial position of the respective elements. Due to reciprocity, the receive beam $p_{r,n}$ can as well be approximated by Eq. 1, but using only a single element as aperture A. Assuming a linear system, the radiation pattern for each element σ_n can be found by multiplying the transmitted beam p_t with the receiving beams $p_{r,n}$:

$$\sigma_n = p_t p_{r,n}. \quad (2)$$

Using Eq. (2), the radiation pattern σ_n is calculated in a plane parallel to A at distance $L = 0.15$ m, which falls into the required working range of the CTG US transducer. In section III the two transmit configurations are compared numerically and experimentally.

B. Fetal beating-heart in-vitro setup

For validation of the method, a dedicated experimental in-vitro setup was designed and realized, visible in Fig. 2. In order to create a simple but sufficiently realistic fetal heart setup, we considered the following requirements to be most important: 1) physical dimensions of the phantom, 2) acoustic properties, 3) possibility for displacement, 4) realistic heart rate and 5) displacement per heart beat.

The biventricular outer diameter of the fetal heart was found to vary between 0.9 cm and 3.9 cm for a gestational age of 14 to 40 weeks [5]. In our phantom, the fetal heart is modelled by a sphere with a diameter $d = 2$ cm. The phantom is made out of polyvinyl alcohol, where 10 micron Al_2O_3 has been added as acoustic scatterer, since this is known for its tissue mimicking acoustical properties [6]. The phantom is threaded onto a thin string to minimize additional reflections from the US beam. Moreover, an acoustic absorber is placed at the back of the water tank to further reduce unwanted reflections.

The possiblity for displacement of the fetal heart out of the ultrasound beam is realized by attaching the US transducer to a translation stage.

A motor brings two connecting plates into motion, which in turn are pulling on the string, such that the heart phantom is drawn towards the transducer surface. A counter weight prevents the heart phantom from floating away and guarantees a stable sinusoidal motion pattern. The adjustable driving voltage of the motor is set such that it simulates the normal average fetal heart rate of 140 beats per minute [7]. The distance between the rotary hinge and the motor was tuned, such that the distance covered by the phantom during one cycle is equal to the distance covered by the fetal cardiac wall during one heart beat, i.e. $d_z \approx 0.5$ mm [8].

C. In-vitro measurements

The seven elements of the US transducer were connected to an US open platform (Vantage 256, VERASONICS, INC., Kirkland, USA). In the performed experiments, when transmitting, either the center element ($n = 4$) alone or all elements were driven. A pulse with center frequency $f_0 = 1$ MHz and pulse duration $T = 20$ cycles was used as driving signal. The pulse repetition frequency was set to PRF = 2000 Hz.

For each of the seven elements, the received raw US data were digitized with a sample frequency of $f_s = 4$ MHz and stored. The directional Doppler Data were obtained by using an IQ-demodulation scheme [9]. Within this demodulation scheme, the sample volume (SV) is selected by integrating over a time window of the raw US Data received per pulse. Therefore, the Doppler signals received in each element have a sample frequency equal to the PRF. Note that the effective sample volume length (SVL) is dependent on the pulse duration T and the duration of the time window $\Delta\tau$ [10], given by

$$\mathrm{SVL} = c_0 \frac{(T + \Delta\tau)}{2}, \tag{3}$$

where c_o is the speed of sound in water. The Doppler frequency f_d is affected by the velocity of the reflector and the angle between the beam and the movement direction of the reflectors within the selected SV. When there is no movement, f_d is zero. If the Doppler signal power P is calculated over a time window W of little heart motion, P will be small. Therefore, it is required to allow several heart beats within the window W. The power is calculates using the root mean square in a window $W = 1$ s. For visualization, P is normalized by the maximal received power during both measurements, color coded and displayed according to the spatial arrangement of the elements, as seen in Fig. 1.

III. Results

Fig. 3a and Fig. 3b show the radiation pattern for the two transmit configurations, derived with the model in section II-A. When transmitting with all elements, strong side lobes are visible. In the presence of side lobes fetal heart location estimation, solely based on the power of the received Doppler signals, is difficult. In contrast to that, transmission with only a single element allows to easily link the received power to an unambiguous fetal heart location. Aiming at a simple method, the modelled radiation pattern suggest to use a single element for transmission.

In the performed experiments the transducer is translated through the water tank, such that the fetal heart phantom was virtually moved along the axis through elements $3, 4$ and 5 at distance $L = 15$ cm, as indicated by the red line in Fig. 1. The received power was measured for the two transmit configurations. Fig. 4a shows the measured normalized power, where all elements were transmitting simultaneously, while in Fig. 4b only the center element, i.e. $n = 4$, was used for transmission.

When transmitting with all elements, clearly the unfavourable effects of the side lobes are visible. Qualitatively the modelled radiation patterns and the measured normalized power for the two transmit configurations match well. Differences between the results from the model and the experiments might be due to misalignment in the experimental setup, the size and the shape of the fetal heart phantom or differences in the transducer elements. Moreover, the power is calculated over $W = 1$ s of the received Doppler signal, in which the location of the phantom has changed slowly. For more comparable results, the received power should be measured while there is no relative displacement

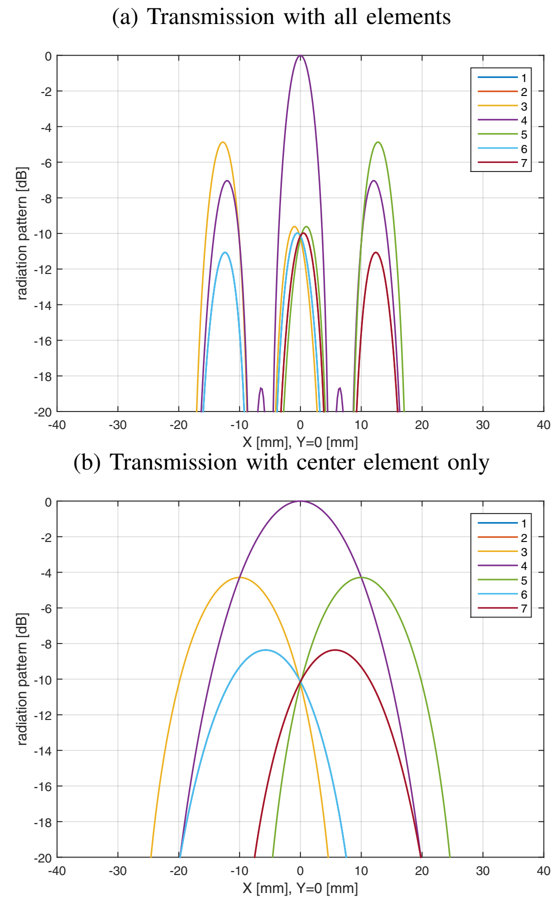

Fig. 3: Cross section of the radiation pattern in the XY plane at $L = 15$ cm and $y = 0$ cm. Note that due to symmetry the radiation pattern of the elements 1 and 2 is equal to the radiation pattern of elements 6 and 7 respectively.

of the fetal heart phantom with respect to the transducer. However, the performed experiment is closer to the situation in clinical practice, where the obstetrician is slowly moving the transducer across the maternal abdomen when searching for the optimal transducer position.

Fig. 5 shows how visual feedback on a monitor could look like. The received normalized P, when transmitting with only a single element, is color coded and shown for each of the transducer elements. The black cross indicates the actual position of the fetal heart during the measurement. With the visual feedback, it can be easily determined when the fetal heart lies centrally in the ultrasound beam and the US transducer can be positioned accordingly.

IV. Discussion and Conclusion

The modelled radiation pattern and the experimental results show, that the problem of ambiguity can be solved by using only a single element for transmission. However, this also reduces the total received normalized power, as shown in Fig. 4b, which may limit the penetration depth of the US wave. In order to detect fetal displacement in a larger volume of observation, the different elements could be used for transmission sequentially.

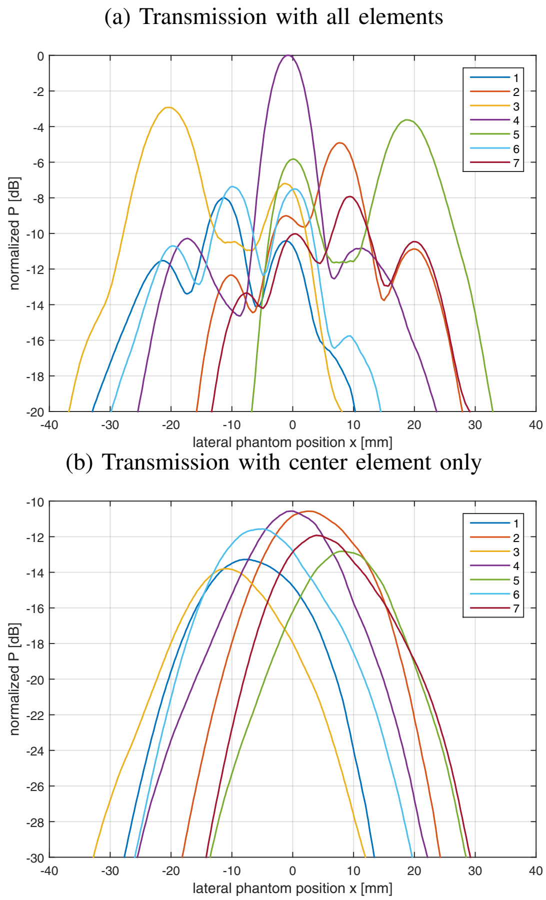

Fig. 4: Measured normalized P. The US transducer is horizontally translated through the water tank. The fetal heart phantom, beating at 140 bpm, is placed at distance $L = 15$ cm and $y = 0$ cm.

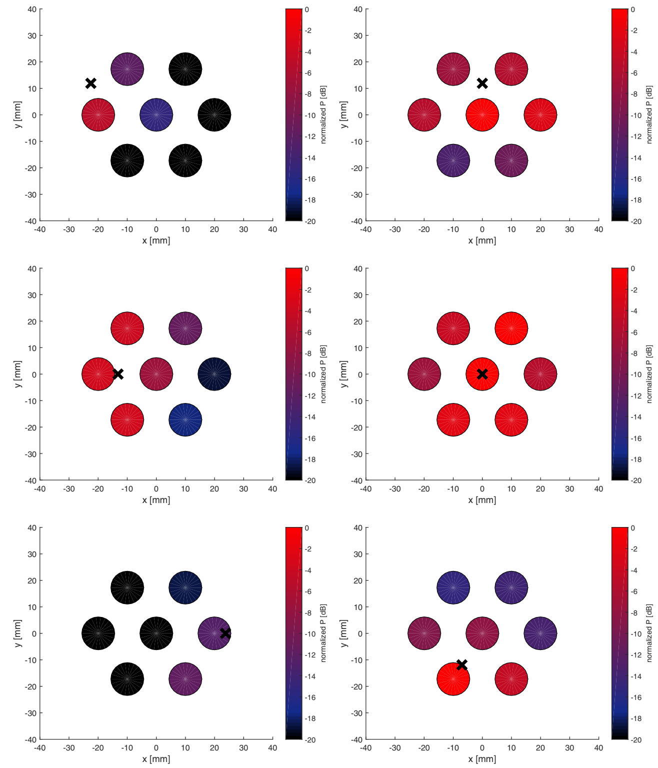

Fig. 5: Visual Feedback for US transducer positioning. The black cross indicates the actual location of the fetal heart phantom at the distance $L = 15$ cm. The colorbar indicates the normalized power P of the received Doppler signals in dB. Only the center element was used for transmission.

In the experiments, the heart phantom was placed at distance $L = 15$ cm from the array aperture, which is in the far field of the US beam. However, in clinical practice the fetal heart can be much closer to the transducer surface. Eq. (2) can only be used to approximate the radiation pattern in the far field. For future work, other models may be used to study the radiation pattern for the entire required measurement range. Furthermore, an algorithm may be developed, which determines the location of the fetal heart more quantitatively.

Additionally, in-vivo measurement will be important to validate the reproducibility of the method in clinical practice.

In conclusion, the results demonstrate the feasibility of obtaining fetal heart location information using the power of the Doppler signals received in the elements of a CTG US transducer. This may aid in transducer positioning and improve the work flow of the medical staff. Possibly, even the mother can reposition the transducer on her own, such that periods of fHR signal loss are directly resolved instead of having to wait for an obstetrician to perform this task.

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