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A test and simulation device for Doppler-based fetal heart rate monitoring

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Abstract: The Doppler effect is the preferred technique in fetal heart rate (FHR) monitoring devices. The main objective of the recent studies on the Doppler FHR has been to improve the accuracy. On the other hand, a reliable fetal heart simulator becomes essential for testing Doppler FHR monitoring devices. The motivation of this study is to design a reliable system that will be used to test Doppler FHR monitors. This device generates a similar Doppler frequency shift of fetal cardiac activity including the heart's wall and valve motions. The components of this system are basically a signal generator using a microcontroller and a modified relay. The relay is the most important and studied component that affects the Doppler frequency shift. Thus, the relay's contact distance and length as well as its closing and opening velocities have been modified to produce an appropriate Doppler shift to fetal cardiac motion. This device has been tested and its reliability has been proved.

Key words: Doppler effect, fetal heart simulation, fetal heart rate, fetal monitoring, ultrasound, quality control

1. Introduction

Since the recognition of fetal wellbeing is crucial during pregnancy, Doppler effect, phonocardiography, fetal electrocardiogram (fECG), and direct scalp ECG methods are commonly used to diagnose fetal wellbeing [1–6].

Usually direct scalp ECG, which is an invasive method, ensures higher accuracy. However, noninvasive methods such as phonocardiography, fECG, and Doppler fetal heart rate (FHR) monitoring provide flexible and portable solutions and do not annoy the patients [7–9]. Since the first study of Doppler FHR in the 1960s [10], many methods and algorithms have been applied to improve the accuracy and remove artifacts of FHR monitoring [11–14]. The artifacts of Doppler FHR monitoring are especially detected by Doppler frequency shifts of fetal movements, breathing movements, and maternal activity that have to be removed or filtered for accurate processing of fetal heart rate signal [15–17].

There are many techniques for extracting the information of the Doppler signal [6], but the operating principle of Doppler FHR monitoring is to detect and calculate the frequency shift between transmitted and received ultrasonic signals [18]. The ultrasound (US) probe of the Doppler FHR monitor is held on the maternal abdominal area to emit and detect the reflected US signal. The placement of the US probe of the Doppler FHR monitor is shown in Figure 1.

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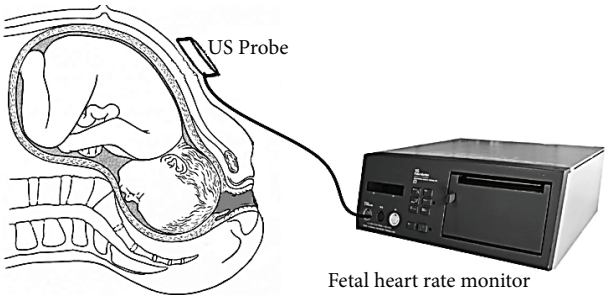

Figure 1. Placement of the US probe on maternal abdominal.

The range of ultrasonic frequency used in Doppler FHR monitoring is between 1 and 2.75 MHz. The US probe emits the ultrasonic wave and receives a reflected US wave from maternal and fetal organs. The frequency shift of the received signal depends on the maternal and fetal activity [19]. The frequency of the shifted waveform is directly proportional to the velocities of fetal breathing movement, fetal global movement, fetal cardiac motions (valve and wall), and maternal activity. To find the Doppler shift, the formula is:

$$f_D = \frac{2vf\cos\theta}{c}, \tag{1}$$

where f is the carrier frequency of the US, v is the target velocity, θ is the angle between the US probe and motion direction of fetal or maternal organs, and c is the velocity of the US signal in soft tissue [20]. Since the velocities of the fetal activities are different, the detected signal should be filtered to process only frequency shift due to fetal cardiac motion including valves and walls motions [21–23]. The useful information in the reflected US beam is found between 100 and 2000 Hz to calculate the periodicity of the fetal heartbeat. The other ranges, especially 0–100 Hz, are related to fetal breathing movement and global movement, which have slower velocities than fetal cardiac motion [24,25].

An FHR monitor and US sensor should be checked for whether they work properly and measure accurately in clinical environment to prevent diagnosing faults. Thus, a reliable testing device becomes an essential requirement for this process. A testing device has to simulate fetal cardiac activity, especially its velocity, in the Doppler system.

The purpose of this study is to investigate the design properties of a device to simulate fetal heart valves and wall motion in air. This device generates similar Doppler frequency shift of fetal heart activity using a modified electric relay in air in order to test Doppler FHR monitoring in a clinical environment. Thus, the relay's opening and closing velocities have been modified to produce similar frequency shifts to the fetal heart in soft tissue detected by an US probe. For the device, a voltage source, an adjustable signal generator using a PIC microcontroller, a common emitter amplifier to drive the relay, and the modified relay have been used. The device has been tested with commercially available Doppler FHR monitors, and its functionality and accuracy are investigated focusing on the relay's effect on the frequency range of Doppler shift.

2. Materials and methods

2.1. Software and hardware implementation

The main function of the Doppler FHR test circuit is to generate adjustable square waveform and drive the relay. For this reason, a variable voltage source, signal generator, and common emitter amplifier are used to make the relay open and closed. The critical point at the hardware and software part of this device is to generate

exact frequency, which is selected by the user turning a gray-coded rotary switch. The generated frequency defines the fetal heartbeats per minute (BPM), found by the formula below:

$$BPM = 60 f_0, \tag{2}$$

where f_0 refers to the frequency of the generated square wave in Hz. The generated frequency controls the timing of the relay's opening and closing sequence, and its physical velocity produces Doppler shift in the reflected US beam. The functional block diagram of the device is given in Figure 2.

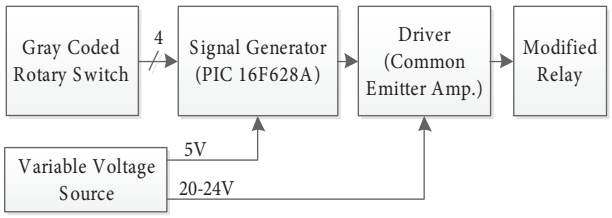

Figure 2. The functional block diagram of the device.

The gray-coded rotary switch gives a 4-bit code for each position and is decoded by PIC microcontroller to generate a square waveform in the range of 1.5–3.3 Hz, which is equal to 90–198 BPM. The follower of the PIC microcontroller is a common emitter amplifier to drive the relay. The variable voltage enables tuning of the closing duration of the relay. When the voltage is adjusted to the maximum level, the relay closing velocity becomes faster.

The software for the PIC microcontroller reads PortA. According to the gray code, it sets the Timer1 module for the required BPM output on PortB. If the exact frequency is not generated due to resolution, a few no operation (NOP) codes are added after timer interrupt. Hence, the exact on-off timing of the square wave is achieved.

2.2. Specifications of the modified relay and Doppler shift

The relay (Panasonic, HC type) is an ordinary electric relay with 24 V working voltage. The difference that makes it modified is the angle between the opened and closed armature points that affects opening and closing velocities of the relay's armature. A schematic representation of the relay is shown in Figure 3.

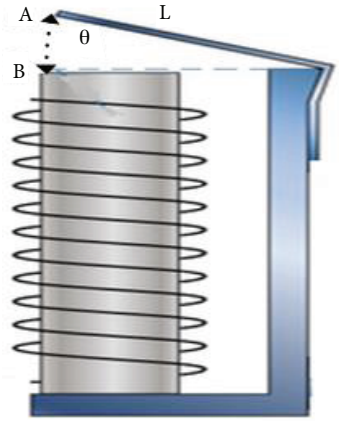

Figure 3. Schematic representation of the modified relay.

L is the length of the armature, and θ is the angle between armature opened (point A) and closed positions (point B). Since the parameters of the relay have significant importance to provide similar Doppler shift to fetal cardiac activity including wall and valve motions, its angle, length, and opening and closing velocities are carefully studied and modified. The parameters of the relay are given in Table 1.

Table 1. The parameters of the modified relay.

L (mm)	$V_{closing}$ (m/s)	$V_{opening}$ (m/s)	θ (°)
7	0.035	0.105	21

The closing and opening velocities ($V_{closing}$ = 0.035 m/s and $V_{opening}$ = 0.105 m/s) in Table 1 were found using an infrared detector module. $V_{closing}$ and $V_{opening}$ differ from each other in the nonmodified relay's opening and closing velocities, while the modified angle, θ, is 21°. The velocity values have been changed to obtain appropriate Doppler shift. Linear motion of the relay's armature causing Doppler shift is assumed as in the model in Figure 4.

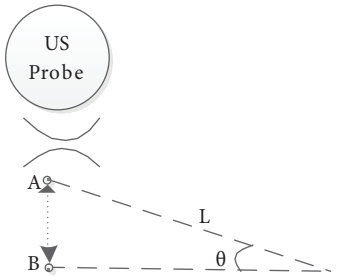

Figure 4. The model of the relay's linear motion.

The maximum linear velocity is measured at maximum length (L = 7 mm). This means that the maximum Doppler frequency shift for 1 MHz US beam during opening ($f_{Dmax}^{opening}$) and closing ($f_{Dmax}^{closing}$) can be calculated using Eqs. (3) and (4) with respect to Eq. (1).

$$f_{Dmax}^{closing} = \frac{2 \times 0.035 \times 10^6}{330} \cong 212Hz \tag{3}$$

$$f_{Dmax}^{opening} = \frac{2 \times 0.101 \times 10^6}{330} \cong 612Hz \tag{4}$$

Since previous studies on Doppler FHR monitoring showed that the useful information is in the range of 100–2000 Hz [22] and much more detailed study giving Doppler shift of fetal cardiac activity proved that the fetal heart's wall motion generates frequency shift in the range of 100–300 Hz and the heart's valves generate frequency shift between 150 and 600 Hz [1]. Both motions of the relay simulate fetal cardiac activities including valve and wall motions in air and generate similar frequency shift to fetal cardiac motion. To summarize and combine the Doppler effects of the system with the circuit and relay device together, Table 2 is given below.

Table 2. The overall device timing and Doppler frequencies in reflected US beam.

Signal waveform	Relay's motion and velocity (m/s)	Estimated max Doppler shift (Hz)	Simulated fetal cardiac motion
⌐	↓ 0.035	212	Wall
⌙	↑ 0.101	612	Valve

Referring to Table 2, the closing movement of the relay starts with the positive edge of the signal simulating the cardiac wall motion; it causes an estimated Doppler shift of 212 Hz. As such, the opening movement triggered by the negative edge of the signal simulates valve motion generating 612 Hz. The frequency of the signal generator controls only the BPM value displayed on the FHR monitor, whereas the physical properties of the relay affect the range of Doppler frequency shift.

3. Results and discussion

Fetal breathing movement, global movement, hiccup movement, and maternal activity produce Doppler shifts, which are generally removed by using a band-pass filter in order to process periodicity of the frequency reflected by fetal cardiac activity. The applied filters and measurements of previous studies describing the Doppler frequency shift by fetal cardiac activity and the Doppler frequency shifts of this device are given in Table 3 to compare.

Table 3. Comparison of previous filtering and measurement frequencies with the simulation frequencies of this study.

Origin	US freq. (MHz)	Filtering or instrumentation
Shakespeare et al. [1]	1.5	Generates 600 & 300 Hz (valve & wall)
Murata & Martin [21]	2	600–2000 Hz for valve motion
Xiaofeng et al. [16]	2.75	1350–2750 Hz for cardiac activity
Jezewski et al. [3]	1	200–1000 Hz for cardiac activity
Foulqure [25]	2.3	5–60 Hz for fetal movement
Maeda [23]	2	20–80 Hz for fetal movement
Jezewski et al. [7]	2	150–350 Hz for cardiac activity
Boss and Schraag [22]	1	100–500 Hz for cardiac activity
Peters et al. [26]	1	100–475 Hz for cardiac activity
This study	1	Simulates valve & wall motion generating 612 Hz & 212 Hz

The investigated useful frequency bands in previous studies are compatible with the simulated Doppler shift in this study referring to Table 3. It can simulate fetal cardiac motions including valve and wall motion in the air. The reflected frequency shifts from the device are in the range of band-pass filtering used to recover fetal cardiac activity and near the frequency shifts of fetal cardiac valve and wall. Since these Doppler frequency shifts depend on the linear velocity of the modified relay, it shows that its modification for changing its armature opening and closing velocities is compatible for simulating frequency shifts of fetal cardiac motion in air instead of soft tissue.

The other critical point in this system is the circuit, especially the square waveform generator using the PIC microcontroller. While testing a Doppler FHR monitor, the user-selected BPM value should be stable to check the accuracy of the US probe or the overall FHR monitor. For this reason, the device has been tested with commercially available FHR monitors (Philips Avalon FM20, HP 50 Series, Corometrics 171). The adjusted BPM and measured BPMs are given in Table 4.

Four of ten testing results have slight error rates between the expected BPM and FHR monitor detection values, which is thought to be caused by the timing properties of the generated square wave or the precision of the FHR monitors. However, the main requirement of this design, compatible Doppler frequency shift in air, is achieved as shown in Table 4. The other measurement results of this device are equal to the detected BPM values. Since the error rate is very small, the designed system can be used as a simulator of fetal cardiac motions for testing a Doppler FHR monitor and its accuracy.

Table 4. Testing results of the device with FHR monitors.

Adjusted apparatus frequency and BPM		Detected BPM by FHR monitor and error rate			
Frequency (Hz)	Expected BPM	HP 50 Series	Philips FM20	Corometrics 171	Max relative error (%)
1.5	90	90	90	90	0
1.7	102	102	101	101	–0.9
1.9	114	114	114	114	0
2.1	126	126	126	126	0
2.3	138	136	136	136	–1.4
2.5	150	151	151	151	0.6
2.7	162	162	162	162	0
2.9	174	174	173	173	–0. 5
3.1	186	186	186	186	0
3.3	198	198	198	198	0

4. Conclusion

Doppler frequency shift has become a commonly used method for FHR monitoring because it is noninvasive and flexible. Since studies focus on the processing of Doppler signal to increase FHR accuracy and because it is necessary to check the accuracy of commercially available FHR monitors, a FHR test device is required. In this study, a Doppler FHR testing device has been developed and its operating principles have been examined. The Doppler frequency shift generated by fetal cardiac motion in soft tissue is simulated by using a circuit that mainly consists of a square waveform generator and a modified relay. The most critical point is the relay motion, which generates appropriate Doppler frequency shift in the air. The modified relay's motion such as the armature's opening and closing velocities in air are adjusted to produce Doppler shift compatible with fetal cardiac motion. According to the difference between the velocity of US in air and in soft tissue, the length of the armature and the angle have been modified. The estimated Doppler frequency shifts of the device have been compared to previous studies and found in the suitable range of the FHR monitor's band-pass filter used to recover cardiac valve and wall motion. Moreover, the difference between the relay's armature opening and closing velocity makes it possible to simulate both Doppler shift of the fetal cardiac valve and the wall in the same period. The device has been tested with commercially available Doppler FHR monitors and only four measurements out of ten have a slight error (max. –1.4%). Thus, its functionality and accuracy have been proved to test the accuracy of a Doppler FHR monitoring device in order to prevent diagnosing faults in clinical environment. Its easy and low-cost implementation can make it a good solution for further applications.

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