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Eddy Current Based Flexible Sensor for Contactless Measurement of Breathing

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Abstract *– Changes in the lung air volume change the conductivity distribution in the thorax. Based on this effect, we describe a non-contact instrument to measure lung volume which is suitable for applications such as monitoring of breathing in sleeping subjects for sleep apnea. The instrument is based on a magnetic coil is positionned in a flexible mattress underneath the subject. AM frequency current is applied to the coil to generates Eddy currents in nearby conductive tissues, which in turn generate a magnetic field opposed to the transmitted field. This effect makes changes in the thoracic conductivity distribution due to breathing and blood flow change the effective inductance of the coil. We describe a new design to measure this effect using a single transmit and receive coil. The coil forms part of a Colpitts oscillator tuned to be at its stability margin, in order to give large frequency changes for given changes in conductivity. The oscillator output is then demodulated using a frequency counter. This device was tested in healthy volunteers and showed a good correlation with pneumotachograph measurements.*

Keywords *– Biomedical Instrumentation, Eddy current monitoring, tissue conductivity*

I. INTRODUCTION

There is significant interest in non-contact monitoring of lung activity, for applications such as monitoring of patients in intensive care for the onset of critical clinical conditions or for monitoring sleeping subjects for apnea. Apnea in infants is associated with sudden infant death syndrome, while adult apnea causes significant difficulties with sleep. While large medical imaging technologies, such as CT, MRI and PET scanning allow accurate non-invasive imaging of the thorax, they are not suitable for use in monitoring applications, due to the bulkyness and expense of the equipment. Many instrumentation technologies also exist for lung function measurement; a good overview is given by [11]. However, very few of these technologies are non-invasive. Perhaps the most widely accepted of these is inductive plethysmography [2], in which inductive bands are placed around the chest and abdomen and the change in diameter of each measured. Inductive plethysmography is currently considered the gold standard for non-invasive lung monitoring in sleep, but has several well known disadvantages [2, 3, 8]: over longer times, the monitoring bands tend to change position as the patient moves, resulting in measurement inaccuracies. Additionally, the placement of wires and bands on a sleeping patient is unconfortable and can induce feelings of claustrophobia.

The requirements of various breathing monitoring applications differ significantly [3]. Some applications, such as infant apnea monitoring, require primarily breath timing. Any significant changes in breathing rate must be detected by the system and result in an alarm. The primary concern is respiratory arrest, or apnea, but other timing changes, such as tachypnea or rapid breathing, should also be detected. Adult sleep apnea monitoring is a more complex application in which ventilatory volumes and flows must be measured in addition to breath timing. This allows apneic events to be classified as to whether they are of obstructive or central origin. Obstructive events arise due to increases in airway resistance, and result in reduced flows for a given pressure generated by the airway muscles. In extreme cases, the airway is blocked and paradoxal ventilation results as the chest and abdomical cavities move in opposite directions. Central apnea, on the other hand, results from a depressed control of ventilation. Thus obstructive apnea typically results in decreased tidal volume, while central apnea results in decreased breathing frequency [3].

We are interested in exploring technology for completely non-contact monitoring based on inductive measurement of thoracic conductivity for lung function measurement. Such an approach has the advantage that it has no cables and straps and can have a more rigid mechanical housing. This technology can monitor heart and lung activity by measuring the conductivity changes caused by the movement of air and blood in the thorax. In order to image the conductivity distribution, techniques such as magnetic impedance tomography (MIT), and electrical capacitance tomography (ECT) have been developed [10]. A series of coils are positioned near the chest, and AM radio frequency signals (in the range 1-10 MHz) are emitted from one coil while other coils measure the signals produced. The RF signals produce Eddy currents in conductive substances in the body (such as the heart and lungs), which, in turn, produce a signal which can be measured. Unfortunately, it appears that such imaging systems are inherently difficult to build, largely because the Eddy current signals are so small compared to the driving signals. Any small movement of the measurement coils produces an artifact much larger than the signals. Thus, according to Griffiths [4], for MIT "no convincing in-vivo images have been successfully produced".

In this study, we are interested in building a system which is less sensitive to sensor position. In order to accomplish this, we do not attempt to calculate an image of the conductivity distribution, but rather a single parameter which correlates to breathing and heart activity. This is somewhat similar to the original Eddy current system of Targan and McFee [9] and extensions to it [7, 12]. The difference between our system and these previous ones is it uses a single coil for both signal transmission and detection, and that this measurement system does not need to be rigid. Specifically, these measurement coils can be embedded into mattress cover which could be placed between the subject and the mattress. The flexible coils are able to tolerate bending due to the subjects weight and movement, and provide a continuous monitoring of the patient breathing pattern.

II. SYSTEM DESIGN

Eddy current based measurement generates radio frequency (RF) current in a coil. This current produces a local magnetic field, which induces current in nearby conductive objects, which, in turn, produces a magnetic field the the opposite direction to the original.

In order to detect Eddy currents, several researchers have build a system with a receiver coil mounted at a fixed alignment to the original [9, 12]. In order to reduce interference from the drive signal, Rossell et al [7] used a planar gradiometer design, while Watson et al. [12] have developed a system which aligns the receiver coil at 90° to the transmitter. Another approach to reduce interference is the use of pulsed eddy current [1]. This approach is well established for use in non-destructive testing of metals for applications such as crack detection in airframes. Unfortunately, since conductivities of biological tissues are much lower than those for metals, the signals from pulsed Eddy current systems are of very low amplitude with very rapid decay in biological applications.

We have designed a system based on a single coil, used for both stimulation and measurement. In such a system, the Eddy currents induce a magnetic field opposite to the one generated in the coil, and give an effective result similar to that of an inductive core. As the conductivity distribution changes within the local magnetic field, the effective inductance of the coil changes. We measure these changes by using the coil as part of an oscillator; changes in inductance thus change the operating frequency of the coil, and the signal is demodulated from the coil output. In order to maximise the signal level, the oscillator is tuned to be at its stability margin. This results in an oscillator with very low Q, but in which the frequency strongly varies with changes in component values.

Since the changes in inductance are small, it is necessary to design sensitive measurement electronics to detect them. Our initial designs were based on AM and FM demodulation of the signals [5, 6]. Unfortunately, the signal levels were sufficiently small that it was necessary to high pass filter the output (using a capacitively coupled amplified with a cut-off frequency of 0.1Hz). This had the undesirable effect of preventing static measurements – which are important for measurement of lung function for these applications.

In this paper, we propose a system design based on a frequency counter measurement of the received signal. Even though conductivity changes introduce small changes in frequency, the required measurement bandwith is low, due to the slow changes in biological signals. Breathing related conductivity changes in humans are limited to about 10Hz. This allows the frequency counter integration period to be relatively long, resulting in more accurate frequency measurements.

Our system produces an RF signal to drive a current into a coil using a 5.8 MHz Colpitts oscillator tuned to be at the margin on stability. Because the oscillator is marginally stable, such impedance changes produce large changes in oscillation frequency. The frequency is measured using an accurate frequency counter. In our system, we use a 10 turn coil of 7 cm diameter, and a measurement period 0.05s, resulting in a frequency resolution of 20 Hz. This measurement frequency corresponds approximately to that of the physiological events of interest.

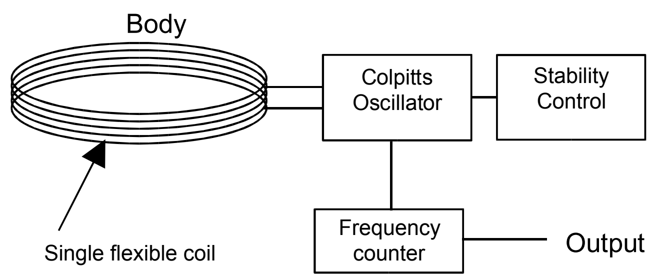

Figure 1: Block diagram of the electronics design. A Colpitts oscillator is designed using a single flexible coil placed on the body surface. Changes in coil impedance due to changes in lung conductivity modulate the coil frequency which are measured with a frequency counter, and the measured frequency is output. In order to maximize circuit sensitivity, the oscillator maintained at the stability margin by stability a control circuit.

III. DATA MEASUREMENTS

In order to test the system of figure 2, we built a system in which the coil was embedded into a foam mattress cover. The volunteer was asked to lie on the mattress and to perform breathing manoeuvres. Air flow was measured using a pneumotachograph connected to a pressure transducer. Data output from the airflow and Eddy current device were recorded using a A/D converter at 20 Hz. A schematic diagram of the data measurement configuration is shown in figure 2. This corresponds roughly to the proposed application of this technology.

Measurements were performed in three healthy male volunteers in the range age 20-40. Experiments lasted approximately 5 minutes, in which time the volunteers were asked to perform several maximum repiratory movements (to total lung capacity and residual volume) followed by a period of tidal breathing.

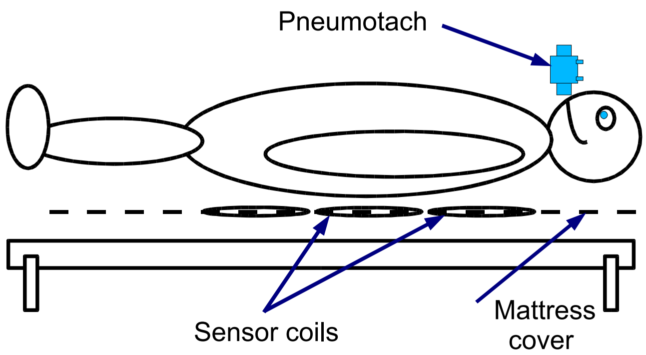

Figure 2: Block diagram of the measurement and test system. The subject lies on a foam mattress into which the measurement coil is embedded. The subject breathes into a pneumotachometer from which air flow is measured using a pressure transducer. The oscillation frequency and flow are then recorded with an A/D converter on a PC.

IV. RESULTS

Data acquired from the volunteers were analysed to determine the correlation between device measurements and pneumotachograph flow. Since the device measures lung conductivity – which is related to volume, we calculate the time derivative of device output to obtain flow. Figure 3 shows sample device output; the upper graph shows the device output (in Mhz) while the center graph shows the pneumotachograph output (arbitrary units). In the data from 0 to 5 seconds, the pneumotachograph output shows expiration. This is expected to result in a decrease in lung air volume, and thus an increase in lung conductivity, resulting in a larger Eddy current load on the oscillator, and a consequent decrease in frequency, as is observed.

Overall, changes in coil frequency are relatively small. The maximum change in frequency between maximal inspiration and expiration was 80kHz. However, the curve of the changes closely follows those measured from lung airflow. In order to compare the device output (which represents lung volume) to the pneumotachograph output (flow), we calculate the derivative of device output, after filtering with a phase neutral low pass filter. This signal is shown on the bottom graph of figure 3. For the first 20 seconds of the signal, there is close correlation between device and pneumotachograph output, while between t=25s and t=30s there is paradoxal movement in opposite directions. We suspect that this is due to motion artifacts in device output.

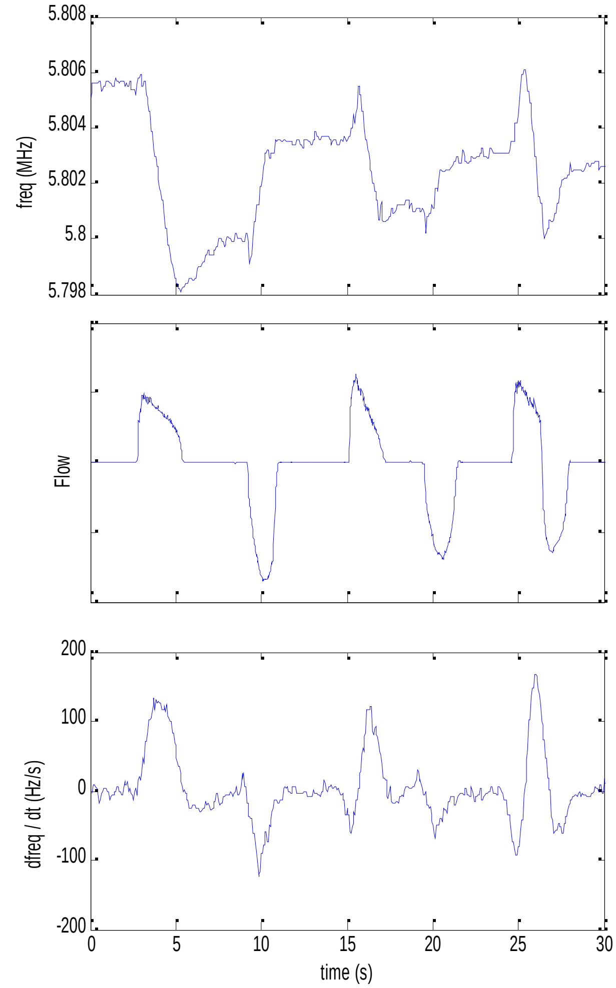

Figure 3: Representative output as a function of time (seconds) from a subject performing forced expiratory and inspriatory breathing manoeuvres with breath holds. *Top*: oscillator frequency (in Mhz). *Middle*: pneumotachograph flow (arbitrary units). Expiratory flow is positive, while inspiration flow is negative. *Bottom*: time derivative of oscillator frequency (in Hz/s).

V. DISCUSSION

We have described a system to perform non-contact monitoring of the breathing activity in a subject. A coil is placed in a mattress under the subject; as the conductivity in the thorax changes, the impedance of the coil changes, inducing changes in the oscillation frequency, which are measured by an accurate frequency counter. Results show the frequency output corresponds roughly to lung volume, and the derivative of the output can be compared to flow measured by pneumotachograph. The advantage of this system is that it is able to use a single, flexible coil for stimulation and measurement.

We are interested in pursuing applications of this technology for monitoring of breathing in sleeping subjects, with application to measurement of sleep apnea. This device has the advantage that it does not require the positioning of wires or straps onto a patient, resulting in less inconvenience and less impact on the patient's normal sleep.

Results of this study are somewhat mixed. The device signal correlates well with pneumotachograph measured flow in a still subject, but can show significant motion artifacts when the subject moves. In the proposed application of sleep monitoring, such motion artifacts are less of an issue; it is sufficient to detect when the patient is moving – it is not necessary to make accurate flow measurements during such movement. We propose to increase the robustness of motion artifact detection by using multiple coil sensors underneath the subject. Movement will be detected differently in each of the sensors, and analysis of these differences in sensor signals should allow motion detection. Additionally, multiple sensors will help ensure a higher signal level if the patient moves away from one coil position on the mattress.

In conclusion, we have described an instrument to make non-contact measurements of breathing. The device is based on a single coil for transmission and reception of RF signals and is thus suitable for applications which require flexible sensors, such as in a mattress cover for sleep monitoring applications.

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