Sensitivity of Single Coil Flexible Electromagnetic Sensors for Breathing Measurements

Rosalyn M. Seeton, BASc, University of British Columbia
Master of Applied Science in Biomedical Engineering Candidate

Thesis Supervisor: Dr. Andy Adler

Ottawa-Carleton Institute for Biomedical Engineering
Department of Systems and Computer Engineering
Carleton University

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Introduction

- Reliable methods of non-contact respiratory monitoring are needed for applications such as sleep and apnea monitoring.
- Electromagnetic sensors provide a non-contact method of measuring changes in conductivity associated with respiratory and cardiac activity but they suffer from low signal strength and are highly subjective to noise.
Background Information

- Commonly used respiratory monitoring techniques
  - Respiratory inductance plethysmography
  - Impedance plethysmography
  - Photoplethysmography
  - Flow sensors (temperature, CO$_2$, humidity, or sounds)

- Methods of non-contact respiratory monitoring
  - Electromagnetic sensors, pressure sensors, ultrasonic sensors, thermal infrared imaging, microwave antenna, CCD camera
Background Information: Electromagnetic Sensors

- Electromagnetic sensors have been developed and studied for magnetic induction tomography or to measure:
  - the conductivity of the head and torso [Tarjan and McFee 1968]
  - changes in brain conductivity [Hart et al 1988][Netz et al 1993]
  - respiratory and cardiac activities [Richer and Adler 2005][Steffen et al 2007]

- Various coil arrangements have been investigated (eg. an excitation coil and one or more receiving coils mounted axially or normally)

- There are different approaches to theoretically model the behaviour of these sensors, but the validation of theoretical models is lacking and the sensor designs are not generally optimized based on the models
Problem Statement

- We seek to investigate the sensitivity of single coil electromagnetic sensors for measuring breathing

Objectives:
- Develop a theoretical model of coil sensitivity to nearby conductivity changes
- Develop a model of oscillator circuit sensitivity from which optimal component values may be selected
- Design and interpret the electrical circuits used as oscillators
- Conduct phantom tests to evaluate the sensitivity of the sensor and validate the theoretical model
- Conduct in vivo tests to measure lung volume changes with the sensor and to compare these to the theoretical model
Theoretical Model: Derivation

- We take the body to be a constant arbitrary shape and we model the lung as a conductive sphere.
- Our current carrying coil is centred below the sphere.
- The theoretical model uses a similar approach as [Hart et al 1988].
Theoretical Model: Derivation

- Use the Biot-Savart law to find the magnetic field

\[ B_z(s, z, N, R) = \frac{\mu_0 N R}{4\pi} \int_0^{2\pi} \frac{(R - s \cos \phi) d\phi}{(s^2 + R^2 + z^2 - 2sR \cos \phi)^{3/2}} \hat{z} \]

- The electric field is found using Faraday’s law

- The current density at each point is then given by Ohm’s law
Theoretical Model: Derivation

- The conductive sphere is broken into very thin disks and summed
- Each disk is modeled as concentric eddy current carrying loops
- The axial magnetic field induced at the plane of the coil by each element of eddy current is then calculated
Theoretical Model: Derivation

- The induced magnetic field causes a change in the magnetic flux inside a loop of the coil giving an electromotive force (EMF)
- The EMF can be seen by the coil as an effective series impedance $\Delta Z$

$$Z = 2\pi\omega^2N(\sigma + j\omega\varepsilon_0)\int_0^R \int_{z_2}^{z_1} \int_{z}^{r} \frac{SS'}{\rho} B_z(s, z, N, R)B_z(s', z, 1, \rho)dsd\rho dzd s' \hat{\phi}$$

$$\Delta Z \propto \omega^2 N^2 \Delta \sigma F_{coil}(R)$$
Theoretical Model: Implementation

- $B_z(s, z, N, R)$ is integrated symbolically
- The rest of the integration is performed numerically

\[
\text{Impedance} = -\omega^2 N \cdot \text{Flux}
\]

\[
\text{Flux} = \int_0^R \int_0^{2\pi} s' \cdot B_{\text{Induced}}(s') d\phi ds'
\]

\[
B_{\text{Induced}}(s') = \int_{z_1}^{z_2} \int_0^{r_z} B(s', z, \rho, 1) \cdot J(\rho, z) d\rho dz
\]

\[
J(\rho, z) = (\sigma + j\omega\varepsilon_0) \cdot E_{\text{Induced}}(\rho, z)
\]

\[
E_{\text{Induced}}(\rho, z) = -\frac{1}{2\pi \rho} \int_0^\rho \int_0^{2\pi} s \cdot B(s, z, R, N) d\phi ds
\]

\[
B(s, z, R, N) = B_z(s, z, N, R) \text{ in elliptic integral form}
\]

- We want to maximize $\Delta Z$
Electronic Design

**Harmonic Oscillator**  **Relaxation Oscillator**
Electronic Design: Colpitts Oscillator

\[ \omega_0 \approx \frac{1}{\sqrt{L \frac{C_1 C_2}{C_1 + C_2}}} \]

\[ \omega_{\Delta Z} \approx \frac{1}{\sqrt{L \frac{C_1 C_2}{C_1 + C_2} \frac{R_1 + \Delta Z}{R_1}}} \]

\[ \frac{\Delta \omega}{\omega} \approx \frac{\Delta Z}{4R_1} \]
Electronic Design: Relaxation Oscillator

\[ f_0 = \frac{R_1}{2L \left[ \ln 2 + \left( \frac{R_1 + R_2}{R_2} \right) \ln \left( \frac{2R_1 + 3R_2}{2R_1 + 2R_2} \right) \right]} \]

If \( R_1 \gg R_2 \), \( f \approx \frac{R_1}{2L \ln 3.3} \)

If \( R_1 = R_2 \), \( f \approx \frac{R_1}{2L \ln \left( \frac{25}{8} \right)} \)

If \( R_1 \ll R_2 \), \( f \approx \frac{R_1}{2L \ln 3} \)
Electronic Design: Relaxation Oscillator

$$f_{\Delta Z} = \frac{1}{2L} \left[ \frac{1}{R_1 + \Delta Z} \ln \left( \frac{2R_1}{R_1 - \Delta Z} \right) + \frac{1}{R_1 \| R_2 + \Delta Z} \ln \left( \frac{\Delta Z}{2R_1} + \frac{2R_1 + 3R_2}{2R_1 + 2R_2} \right) \right]$$

If $R_1 \gg R_2$, $f \approx \frac{R_1 + \Delta Z}{2L \ln 3.3}$

If $R_1 \ll R_2$, $f \approx \frac{R_1 + \Delta Z}{2L \ln 3}$

If $R_1 = R_2$, $f \approx \frac{R_1 + \Delta Z}{2L \ln \left( \frac{25}{8} \right)}$

$$\frac{\Delta f}{f} \approx \frac{\Delta Z}{R_1}$$
Electronic Design

Colpitts Oscillator
- Maximize l, d, N
- Minimize $R_1$, C
- Best R at R~4.5cm

Relaxation Oscillator
- Maximize $R_1$
- Minimize N, R

Coil
- Chose R=4.6cm, N=10
Experiments: Phantom Trials
Experiments: Phantom Trial #1

**Colpitts Measured**

- 0.01 S/m
- 0.61, 0.59 S/m
- 0.98, 0.93 S/m
- 1.33, 1.31 S/m
- 1.88, 1.85 S/m

**Relaxation Measured**

**Colpitts Theoretical**

**Relaxation Theoretical**
Experiments: Phantom Trial #2
Experiments: Phantom Trial #3

- Colpitts Measured
- Relaxation Measured

- Colpitts Theoretical
- Relaxation Theoretical

Relative Change in Frequency $\Delta f/f$

Volume of fluid (mL)
Experiments: In Vivo
Experiments: In Vivo
Summary of Contributions

- A theoretical model was developed to predict the changes that a single coil sensor would experience from conductivity variations of the thorax caused by respiration.
- Several oscillator circuits were investigated, two circuit designs were chosen, analysed, and modeled in terms of relative change in frequency, and components were selected to optimize the sensitivity.
- The theoretical electromagnetic and electronic models were tested with phantom trials.
- The feasibility of using a single coil sensor to measure changes in lung volume was tested with in vivo trials.
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Theoretical Model: Implementation

\[ \frac{\Delta Z}{\omega^2 N^2 \Delta \sigma} \] versus \( R \)
Electronic Design: Relaxation Oscillator
Experiments: In Vivo

- The averages and standard deviations of frequency changes from the lung manoeuvres:

<table>
<thead>
<tr>
<th></th>
<th>TLC</th>
<th>1 Litre</th>
<th>400 mL</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\bar{x}$</td>
<td>$\sigma$</td>
<td>$\bar{x}$</td>
</tr>
<tr>
<td><strong>Colpitts</strong></td>
<td>469</td>
<td>210</td>
<td>124</td>
</tr>
<tr>
<td>$\Delta f$ (Hz)</td>
<td>203</td>
<td>91</td>
<td>54</td>
</tr>
<tr>
<td>$\Delta f \cdot 10^6$</td>
<td>869</td>
<td>698</td>
<td>280</td>
</tr>
<tr>
<td><strong>Relaxation</strong></td>
<td>3517</td>
<td>2820</td>
<td>1139</td>
</tr>
<tr>
<td>$\Delta f$ (Hz)</td>
<td>203</td>
<td>91</td>
<td>54</td>
</tr>
<tr>
<td>$\Delta f \cdot 10^6$</td>
<td>869</td>
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- The probability that the measured frequency changes could be explained by noise:

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</tr>
</thead>
<tbody>
<tr>
<td><strong>Colpitts</strong></td>
<td>5.34$\cdot 10^{-9}$</td>
<td>8.7$\cdot 10^{-3}$</td>
<td>2.26$\cdot 10^{-5}$</td>
</tr>
<tr>
<td><strong>Relaxation</strong></td>
<td>5.99$\cdot 10^{-5}$</td>
<td>6.29$\cdot 10^{-2}$</td>
<td>5.41$\cdot 10^{-5}$</td>
</tr>
</tbody>
</table>