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A Thoracic Model using Three-dimensional Finite Element Method

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= **Abstract** =

A three-dimensional thoracic model was constructed using 8-node trilinear hexahedron elements. A three-dimensional steady-state finite element code was developed using FORTRAN. Its output consists of potential at each node, current in each element, and total current in each layer in the z-direction. The thoracic model was implemented to calculate basal impedance(Z_0) in Impedance Cardiography. Generalized Laplace's equation was solved with Dirichlet(constant potentials) and homogeneous Neumann(no flux) boundary conditions. It was found that the constructed thoracic model was reasonable since the calculated potential differences between the adjacent electrodes and basal impedance were about the same as the measured ones.

1. Introduction

In impedance cardiography, constant current between 20 and 100 KHz at 0.1 to 4 mA is applied to an outer pair of metallic electrodes as shown in Fig. 1. One of which is either around the neck or on the surface of the forehead and the other is around the abdomen. The voltage is measured between the pair of inner electrodes. If the impedance of the thorax changes for any reason, it will cause a change in the voltage measured across the inner electrodes. This change in voltage, which is proportional to the change in impedance, is recorded and used to estimate changes in blood volume.

Impedance cardiorgraphy has been used extensively for monitoring stroke volume because it is noninvasive, has good reproducibility, gives beat-by-beat information, and is simple. However the sources of the measured impedance change and their relative contributions to the thoracic impedance change are incompletely understood even though the values of cardiac output measured simultaneously by impedance cardiography and invasive clinical methods such as thermodilution correlate highly. For complete comparative studies of cardiac output, please refer to the appendix of Mohapatra [1]. Due to this lack of scientific basis, impedance cardiography has not been widely accepted by clinicians despite the advantages described above.

Considerable empirical studies have been done to identify the sources of the impedance change, but the data obtained from animal experimentation must be extrapolated with care to human measurements because of the different thoracic geometries between humans and animals. Hence the purpose of this study is to construct an

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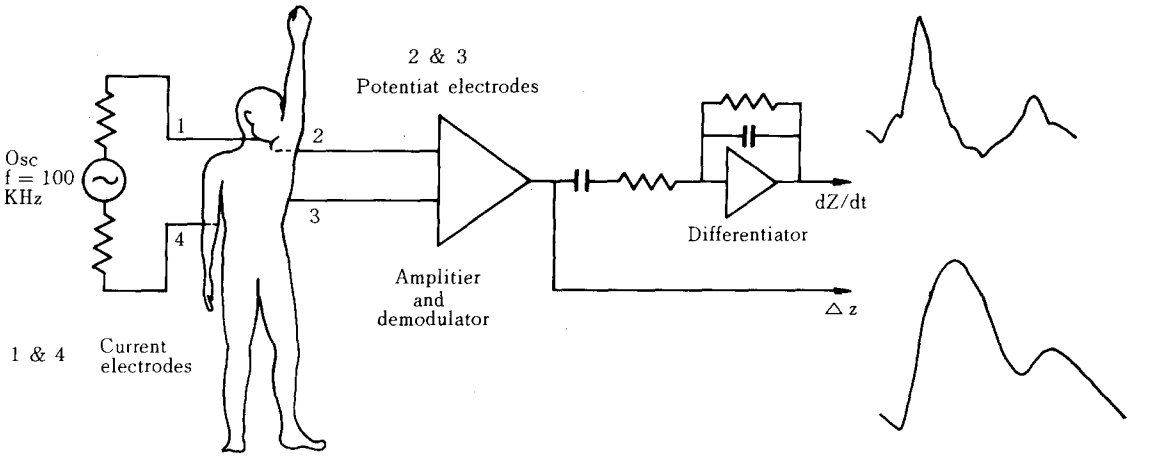

Fig. 1. Schematic diagram of impedance cardiography.

accurate thoracic model in order to investigate the sources of the impedance change and their contributions for future study. More accurate stroke volume may be obtained based on this investigation.

2. Assumptions

For every numerical model, some assumptions are necessary and those assumptions should be reasonable. Otherwise the results will be inaccurate or totally meaningless. The following assumptions were made as compromises between available computing resources and information on tissue properties and thoracic anatomy.

1) Most tissues were assumed isotropic. However, isotropy was not assumed for blood. Even though skeletal muscle shows considerable anisotropy [2], it is very difficult to assign longitudinal and transverse resistivities for each element because of the complicated orientations of the skeletal muscles especially in the thorax. Thus it was assumed that muscle is isotropic.

2) Muscles were assumed to have a relative dielectric constant of about 120 at 100 kHz [Pearce, private communication] and a conductivity of about 0.3 S/m at 100 kHz, therefore the loss tangent is approximately 450. Thus muscles may be assumed essentially resistive at the frequency of 100 kHz.

3) The thorax was assumed to be a source-free region. Even though there are several bioelectrical signal sources produced by the heart (ECG), muscle (EMG), and brain (EEG) etc., these sources do not interfere with the modulated current at 100 kHz, which is much higher than the bioelectric signal sources.

4) It was assumed that there is no current flux from inside of the body normal to the body surface. As the ratio of the body conductivity to that of air is almost infinite this assumption can be made safely.

5) Since we wish to approximate the 100 kHz oscillation of current and potential by constant values, we must be sure the magnetic field is insignificant. It was shown by Kosciki [3] that the magnetic field produced by the 100 kHz current source is negligible.

6) From Assumption 3) there is no electrical source inside the body. Thus the governing equation of this study is Laplace's equation. Then

$$\nabla \cdot J = 0$$

$$-\nabla \cdot (\sigma \nabla V) = 0$$

where J is current density and V is potential.

$$\partial / \partial X(\sigma_x \partial V/\partial X) + \partial/\partial y(\sigma_y \partial V/\partial Y) + \partial/\partial Z(\sigma_z \partial V/\partial Z) = 0$$

where σ_x, σ_y, and σ_z are conductivities of material in the x, y, and z directions respectively. This is the generalized form of Laplace's equation and can be solved subject to the following boundary conditions: a) All points of the body surface in contact with an electrode

have the same potential as the electrode. b) At all other surface points the gradient normal to the surface will be zero. The solution of this boundary value problem is unique as the boundary conditions are given as constant potentials and no flux for all boundaries. Since the geometry of the thorax is complicated, an analytical solution is not possible.

3. Model Description

As the geometry of the human thorax is so complex, a three dimensional model is needed to obtain accurate results. However the construction of a three-dimensional model is quite laborious because of the difficulty in visualizing the thorax in three dimensions and because of the large number of irreularly shaped structures.

Table 1 shows the resistivity assigned for each organ. Except for muscle, thoracic wall, trachea, air + muscle, and air, the tissue resistivities in Table 1 were taken from Geddes and Baker [2] . The thoracic wall is composed of muscle and fat. The resistivity of air is almost infinite but it was reduced to 10^5 to minimize roundoff error in the matrix calculation.

Table 1. The tissue resistivities used in this model

Tissues	Resistivities (Ω -cm)
Muscle	300
Thoracic Wall	400
Blood	150
Fat	2000
Lung	1200
Bone	2000
Intestine	800
Liver	625
Kidney	600
Spleen	300
Stomach	800
Heart muscle	450
Trachea	2000
Cerebrospinal fluid(CSF)	63
Air + Muscle	10^4
Air	10^5

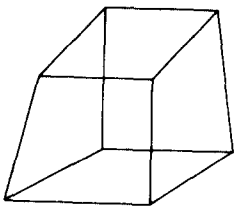

Fig. 2. Eight-node trilinear cubic element.

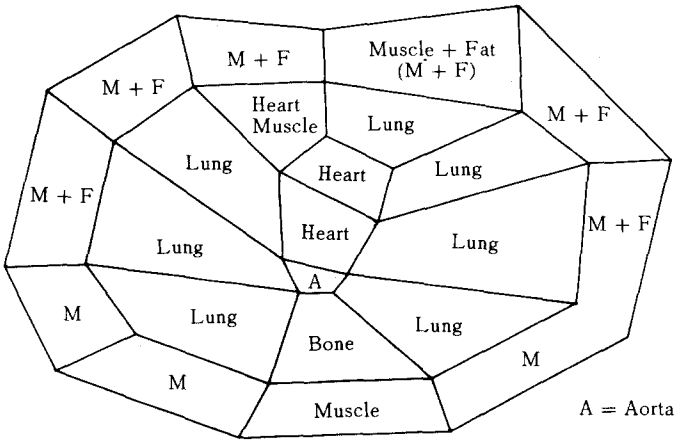

Fig. 3. A typical grid at the level of heart(G10 in Fig.4).

The model of the thorax and neck was constructed from horizontal crossl-sections taken from anatomical maps [4] using eight-node trilinear cubic elements having top and bottom surfaces in parallel as shown in Fig. 2.

The three-dimensional model consists of 22 elements on each of 29 layers. Fig.3 shows the grid of a typical cross-section. The layers were parallel to the x-y plane and nonuniformly spaced for a total of 880 nodes and 658

elements. The constant potentials, 202 mV and 0 mV, measured between the current electrodes on the author using a conventional impedance cardiograph were assigned to the current electrodes to drive the solution.

Each circumferential potential measuring electrode was simulated by ten thin elements having high conductivities of 100 S/m. Thus a layer containing a potential electrode has 10 elements in addition to the 22 elements of other layers. The height of the model was 47.5cm, extending 5cm beyond each current electrode. The widths of the potential and current electrodes were 0.7cm and 1.2cm respectively. The distances between the potential and the current electrodes on the neck and on the abdomen were 4cm and 6cm respectively. Fig. 4 shows the levels of the 29 layers.

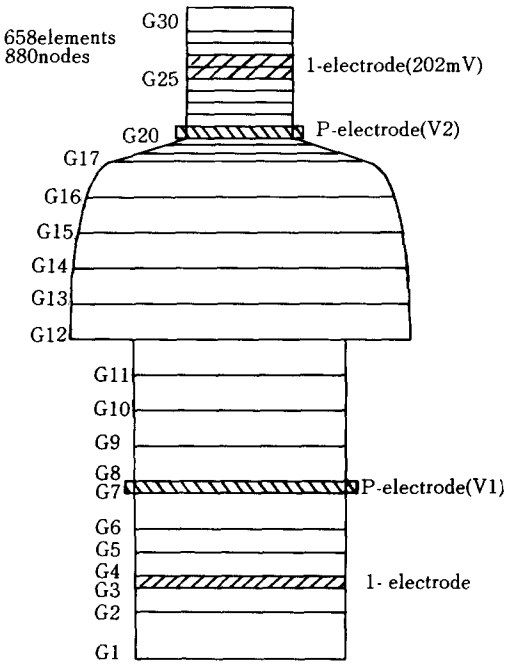

Fig. 4. The levels of the 29 layers of the model.

To validate the accuracy of the constructed computer model of the thorax and the neck, some parameters are needed so that the measured and calculated values can be compared. For a valid comparison the model and subject should have approximately the same physical dimensions and tissue resistivities. The dimensions of the model were about same as those of the subject. The parameters considered for investigating the accuracy of the model were predicted potentials on the potential electrodes on the neck and thorax(V_2 and V_1 respectively), the potential difference(V_2-V_1), total current(I), and basal impedance(Z_0).

Table 2. The measured and calculated parameters

	Measured	Calculated	Error(%)
V_2(mV)	149.0	141.76	-4.9
V_1(mV)	35.0	30.89	-11.7
$V_2 - V_1$(mV)	114.0	110.87	-2.7
I(mA)	4.0	3.62	-9.5
Z_0(Ω)	28.5	30.63	7.5

Table 2 shows that all the calculated values are within an acceptable range of the measured values when the published resistivities of the organs [2] as shown in Table 1 are used. Considering the nature of this numerical modeling of the human thorax, the constructed model can be assumed reasonable.

4. Discussion

In this study, a three-dimensional human thoracic model was constructed and it was verified that the model is reasonable. In order to construct an accurate model, both the resistivity assigned for each organ and the geometry of the model should be accurate. However the published resistivities of the organs were obtained in-vitro are different from the actual resistivities in-vivo. Approximately half of the body consists of muscle [5]. Thus the resistivity assignment for the muscle is crucial for an acc urate thoracic model.

Using this model the sources of the impedance change due to physiological changes during the cardiac cycle can be investigated by simulating each change and calculating the impedance change. Once the contribution of each source to the impedance change is known, it may help measure more accurate stroke volume and provide scientific basis for impedance cardiography.

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국 문 요 약

3차원 유한 요소법을 이용한 흉부 모델

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본 연구에서는 8-node cubic element 로서 3차원 흉부모델을 구성하였고 또한 3차원 정상상태 유한요소법 코드를 FORTRAN을 이용하여 개발하였다. 프로그램의 출력은 각 node의 전위, 각 요소의 전류, 그리고 Z-축방향으로의 각 층에서 총 전류이다. 이 흉부모델을 이용항 임피던스혈량측정법에서의 기초임피던스 (Z_0)를 계산하였다. 일반화된 라플라스 방정식을 정전위와 무전속 경계조건을 적용하여풀었다. 또한, 전극간의 전위차와 Z_0의 계산값이 실측치와 근접하였으므로 이 모델이 인체의 흉부와 유사함을 확인하였다.

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