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Optimization of the electrode configuration of electrical impedance myography for wearable application

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ABSTRACT

Electrical Impedance Myography (EIM) based on the four-electrode method is a novel method for assessing muscle state in the fields of sports, fitness, and medical rehabilitation. However, commonly used configuration of electrodes is not suitable for the wearable field, because of its large total area and low sensitivity. An optimized electrode configuration for wearable application is proposed as Mode B. Equivalent circuit model B of the four-electrode method is established by using the equivalent circuit of biological tissues, and in-vivo measurements of the electrical impedance of the biceps muscle are carried out on six volunteers using bioimpedance spectroscopy device Imp™ SFB7. The experimental results show that equivalent circuit model B of the four-electrode method is reliable. Moreover, the variation in muscle electrical impedance measured using the optimal configuration of model B is twice that measured using the optimal configuration of model A. The optimized electrode configuration of EIM based on this approach is model B (i.e. square electrodes in parallel array; size, 20 mm × 20 mm; spacing, 5–24 mm).

1. Introduction

Nowadays, muscle fatigue has become a common symptom in daily life, and has caused widespread concern in the fields of rehabilitation medicine, exercise mechanics, and ergonomics. Muscle fatigue is an important motor characteristic of muscle tissue [1] and is generally reversible. If it is not treated in time, it might lead to permanent damage, which can have a serious impact on people's work, exercise and daily life [2]. Therefore, it is of great significance to effectively assess and monitor the fatigue degree of muscle tissue at any time and place in the field of exercise, fitness and rehabilitation medicine, which can reduce the damage caused by exercise or other muscles caused by muscle injury.

In the field of wearable muscle fatigue detection, the technique based on Electrical Impedance Myography (EIM) has gained increased attention due to its miniaturization, non-invasive, convenience and realisability of anytime, anywhere detection. EIM is a new method for detecting and evaluating muscle diseases [3], which has been clinically verified [4]. EIM muscle fatigue detection technology is a common and non-invasive four-electrode impedance measurement method, which consists of a pair of excitation electrodes and a pair of measuring electrodes. The measurement

principle is to apply high-frequency and low-intensity current (I) to the experimental muscle mass or muscle group, and evaluate the fatigue state of the muscle by measuring the amplitude and phase of the coupling voltage (V) [3]. Explained from the energy loss, due to the inherent resistance characteristics of biological tissues, the current flowing into the biological tissues will cause losses, resulting in a reduction in the voltage amplitude at the receiving end; at the same time, because of the capacitive characteristics of biological tissues, the voltage signal at the receiving end delays occur. According to Ohm's law, the muscle electrical impedance R is calculated by Equation (1). The muscle electrical impedance R changes when muscle fatigue or muscle properties change. These variations of muscle electrical impedance ΔR detected by EIM muscle fatigue detection technology provide an assessment of the progress or relief of muscle fatigue.

$$R = V/I. \tag{1}$$

EIM relies on local current flow through muscle tissue, so the electrode configuration of EIM remarkably affects the measurement results [5]. Electrode configuration involves electrode size, spacing and arrangement. Jafarpoor [6] and Somen [7] optimized the electrode configuration and determined the optimal electrode
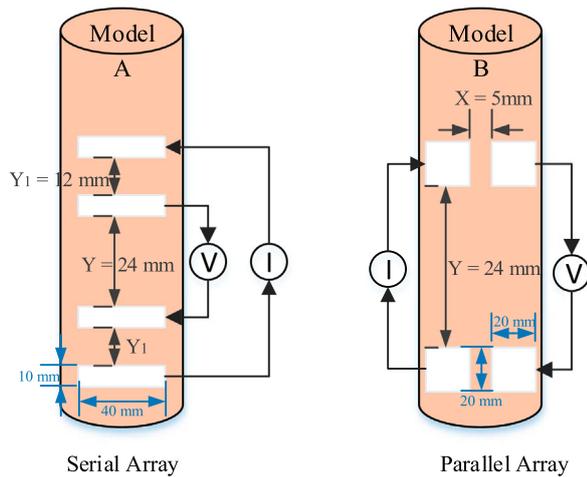

Figure 1. Electrode configurations of EIM based on the four-electrode method. Model A represents the traditional four-electrode configuration with a serial array. Model B represents the novel four-electrode configuration with a parallel array.

configuration by means of modelling and simulation, in which the excitation (outer) electrodes and the measuring (inner) electrodes are arranged in line along the z-axis, as shown in Figure 1 (model A). In existing studies, researchers use model A of the electrode configuration to detect and evaluate muscle diseases and state [8]. However, since model A configuration has large area and low potential sensitivity, it is not suitable for the miniaturization and portable design of wearable devices. Therefore, in order to further optimize the electrode configuration for detecting and evaluating muscle disease in the wearable field, we adapted the electrode arrangement to make the array more compact and have higher potential sensitivity.

In the electrode arrangement proposed in this paper, two pairs of square electrodes are arranged symmetrically along the z-axis, as is shown in Figure 1 (model B). And square electrodes with the same area are selected to make the array more compact in Model B. A new equivalent circuit model of the four-electrode EIM method has been proposed by analysing equivalent circuit model of the four-electrode method, based on the equivalent circuit of biological tissue. *In vivo* measurements of distribution of electric current measured on biceps muscle in the resting state and during muscle fatigue have been conducted using bioimpedance spectroscopy device Imp^{TM} SFB7. The developed equivalent circuit model B is verified, and the applicability of model B to replace model A configuration for detecting muscle fatigue in wearable devices is discussed. Variations in muscle electrical impedance (ΔR) during dynamic motion are also analysed. Finally, an optimal electrode configuration of EIM based on the four-electrode method is proposed for muscle fatigue detection in the wearable field.

This paper is divided into the following sections. The research background and purposes of the study

are introduced in Section 1. Equivalent circuit model B based on the four-electrode method is proposed in Section 2. *In vivo* experiments with bioimpedance spectroscopy device Imp^{TM} SFB7 are presented in Section 3. The experimental data are analysed in Section 4. Finally, the conclusions of this study are drawn in Section 5.

2. Equivalent circuit models

The four-electrode method of EIM for muscle fatigue detection consists of a pair of excitation electrodes and a pair of measuring electrodes. This approach can effectively eliminate the contact impedance produced by contact between the electrodes and muscle tissue to improve the accuracy of the measurement system [9,10]. Moreover, the measurement principle can detect changes in muscle electrical impedance by applying an excitation current signal in response to a voltage signal. The measurement principles of conventional Model A and proposed Model B are essentially identical. Both models are based on the four-electrode method of EIM, as presented in Figure 1. Model A represents the conventional four-electrode configuration with a serial in-line arrangement in which the excitation and measuring electrodes are connected in series along the z-axis. Model B represent the new four-electrode configuration with a parallel arrangement in which the two pairs of electrodes are aligned symmetrically along the z-axis. In order to improve the density, four equally sized square electrodes were used in this study in Model B experiments.

In Figure 1, Y_1–Y–Y_1 represents the spacing between successive electrodes in Model A, where Y is the distance between positive and negative measuring electrodes and Y_1 is the distance between excitation and measuring electrodes. X–Y represents the spacing of electrode configuration in Model B, where X is the distance between excitation and measuring electrodes and Y is the distance between positive and negative electrodes of the excitation or measuring electrode pair.

The principle of EIM based on the four-electrode method for muscle fatigue detection is analysed in detail according to the three-element model of biological tissue [11–13]. The three-element model is considered to be connected in series by the resistance of intracellular fluid R_i and the capacitance of cell membrane C_m, and then parallel with the resistance of extracellular fluid R_e according to dielectric properties of tissues [14]. Regardless of the serial or parallel array of the four-electrode method, both circuit systems adopt the excitation mode of a constant current source. The excitation electrodes are denoted 1 and 4, while the measuring electrodes are denoted 2 and 3. Z_1, Z_2, Z_3, and Z_4 are the contact impedances produced by the contact between the electrodes (1, 2, 3, and 4) and muscle tissue in the theoretical models of the four-electrode
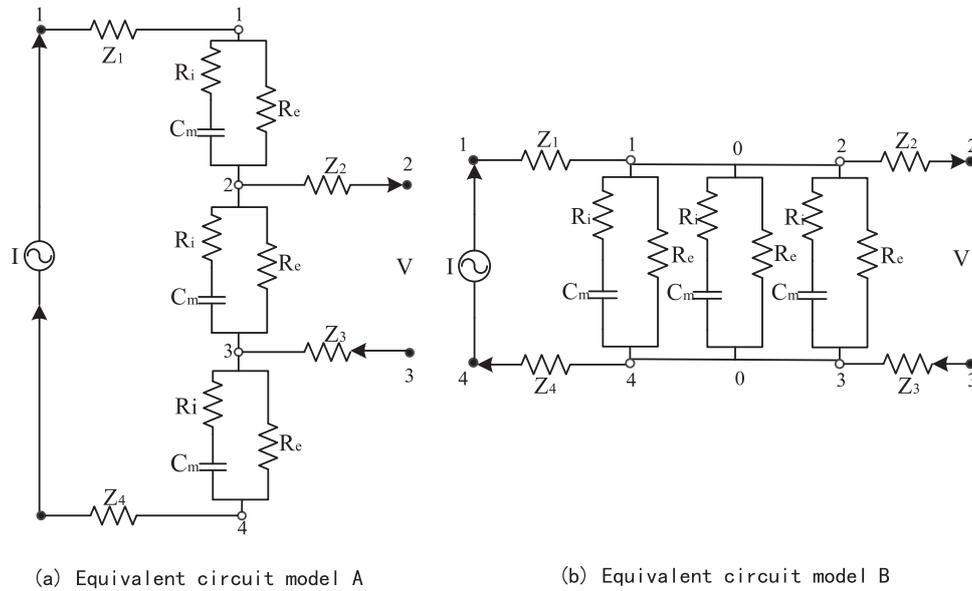
(a) Equivalent circuit model A   (b) Equivalent circuit model B

Figure 2. Equivalent circuit model of the four-electrode method. Model of the four-electrode method in serial array is (a) equivalent circuit model A, model of the four-electrode method in parallel array is (b) equivalent circuit model B.

method. Figure 2(a) illustrates equivalent circuit model A of the four-electrode method in serial array, while Figure 2(b) demonstrates equivalent circuit model B of the four-electrode method in parallel array.

In the existing equivalent circuit model A of the four-electrode method in serial array, the measured muscle tissue can be regarded as a serially arranged group of three small segments of muscle tissue divided as Z_{12} between electrodes 1 and 2, Z_{23} between electrodes 2 and 3, and Z_{34} between electrodes 3 and 4. And each small segment of muscle tissue is equivalent to a three-element mode. The equivalent circuit of model A is shown in Figure 2(a), where I is the excitation current signal of the system, V is the response voltage signal of the tested muscle tissue, R_i and R_e are the resistances of intracellular and extracellular fluids, respectively, and C_m is the capacitance of cell membrane. Z_1, Z_2, Z_3, and Z_4 are the contact impedance values produced by the contact between the electrodes (1, 2, 3, 4) and muscle tissue.

Given that the current flowing through the muscle tissue is evenly distributed, i.e. $I = I_{12} = I_{23} = I_{34}$, the output voltage avoids the influence of the contact impedances of Z_1 and Z_4. The output voltage V is measured by an amplifier with a high input impedance, and the current flowing into the amplifier is close to 0. Hence, the output voltage is unaffected by the contact impedances of Z_2 and Z_3. The traditional electrode configuration can effectively eliminate the interference introduced by contact impedances to ensure that the output voltage is only related to Z_{23} and the excitation constant current source I [10]. In summary, the muscle electrical impedance measured by traditional equivalent circuit model A is shown by Equation (3).

$$Z_{23} = \frac{V}{I}, \tag{2}$$

where

$$I = I_{12} = I_{23} = I_{34}. \tag{3}$$

This study proposes a new equivalent circuit model B in parallel array based on the three-element model and equivalent circuit model A of four-electrode method, as shown in Figure 2(b) where Z_1, Z_2, Z_3, and Z_4 represent the contact impedances produced by the contact between the electrodes 1, 2, 3, and 4, respectively and muscle tissue; I denotes the excitation current signal of the system; V is the response voltage signal of the tested muscle tissue; R_i and R_e refer to the resistances of intracellular and extracellular fluids, respectively; and C_m represents the capacitance of the cell membrane.

First of all, let's assume that the measured muscle tissue can be considered a parallel-arranged group of three small segments of muscle tissue, divided as Z_{14} between electrodes 1 and 4, Z_{23} between electrodes 2 and 3, and Z_{00} in the middle segment. Then, the excitation current of the system expressed by Equation (4), is the current applied into the tested muscle tissue from electrode 1 to 4. According to the same principle as for equivalent circuit model A, this new mode is good to avoid the effects of the contact impedances Z_1 and Z_4 generated by the excitation electrodes and tested tissue, respectively. The output voltage V is measured by an amplifier with a high input impedance, and the current flowing into the amplifier is close to 0. Therefore, the output voltage V is not influenced by Z_2 and Z_3, which are generated by the measuring electrodes and tested tissue, respectively. Thus, equivalent circuit model B based on the four-electrode method in parallel array can effectively eliminate the interference generated by contact impedance, similar to equivalent circuit model A. The overall impedance Z measured by
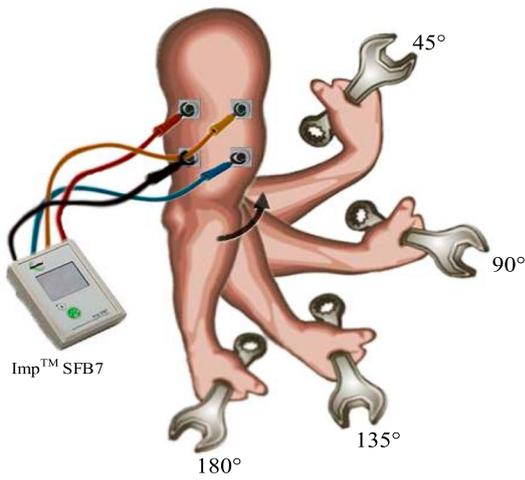
Figure 3. *In vivo* muscle fatigue movement experiment of biceps brachii with Imp^{TM} SFB7.

equivalent circuit model B is shown by Equation (5). According to Equation (6), the following hypotheses can be put forward to further verify equivalent circuit model B through *in vivo* experiments: when the distance between the positive and negative electrodes Y is long, that is, when the measurement length L is large, the resistance R_S is large; when the distance between the excitation and measuring electrodes X is long, that is, when the cross-sectional area S is large, the measured R is small.

$$I = I_{Z_1} = I_{Z_2} = I_{Z_{14}||Z_{00}||Z_{23}}, \quad (4)$$

$$Z = Z_{14}||Z_{00}||Z_{23} = \frac{V}{I}, \quad (5)$$

$$R_S = \rho \frac{L}{S}. \quad (6)$$

3. Method

This research has three objectives as follows: (1) verify the correctness of equivalent circuit model B according to the hypothesis proposed in the static experiment; (2) discuss the applicability of replacing model A with model B configuration for wearable muscle fatigue detection according to the muscle electrical impedance R in different states of fatigue; (3) optimize the electrode configuration and analyse ΔR during dynamic movements to improve potential sensitivity. In order to explore these three research objectives, *in-vivo* experiments were conducted on six volunteers. The experiments are divided into static and muscle fatigue movement experiments, as shown in Figure 3.

The optimal electrode configuration of Model A for measuring the electrical impedance of the biceps muscle was investigated through finite element simulations by our research group [18] and the proposed configuration is shown in Figure 1, left. The size of the electrodes is 10 mm × 40 mm and spacing between successive electrodes (Y_1–Y–Y_1) is 12 mm–24 mm–12 mm. In

order to eliminate the measurement impact caused by the inconsistency between the material and size of the electrode area [15], square electrodes with size of 20 mm × 20 mm were used in model B configuration, as in Figure 1, right. A bioimpedance spectroscopy device Imp^{TM} SFB7 was used for muscle fatigue detection based on EIM.

Six healthy volunteers have taken part in the *in vivo* experiments, including 2 girls and 4 boys aged 22–24 years and weighing 47–80 kg. The volunteers have no history of muscle injury, cardiovascular disease, or upper extremity disease. All of them ceased intensive muscle training at least one week before the beginning of the experiment. Each volunteer understood the entire experimental process and volunteered to participate in this study.

3.1. Static experiments

The first set of *in vivo* experiment aims to explore the relationship between the different electrode configurations of Model B and muscle electrical impedance by X and Y when the biceps muscle is in static state. During the measurements, the upper arm is in a natural droop state, so the angle between the upper and lower arms is 180°. The ultimate goal of the static experiments is to verify the correctness of theoretical model B and determine the optimal configuration of model B electrodes. A bioimpedance spectroscopy Imp^{TM} SFB7 was used to measure the response voltage of biceps muscle by inputting a 2 mA current signal between the excitation electrodes and sweeping its frequency between 4 kHz and 1 MHz. Two sets of static experiments were performed. In the first set, measuring electrodes were placed on a biceps muscle separated by $Y = 24$ mm, and excitation electrodes were moved from the position $X = 5$ mm to $X = 20$ mm with 5 mm step. In the seconds set of the static experiments, distance X was fixed to 5 mm, and distance Y was changed between 5 and 45 mm with a 5 mm step.

3.2. Experiment on muscle fatigue during movement

The second set of *in vivo* experiments involves measurements of muscle impedance measured using optimal model A and model B configurations during dynamic contractions (lower arm flexion, as in Figure 3). In the initial state the volunteers were asked to keep the arm in the state of natural sagging (180°). One movement cycle is change of the angle between the upper and power arm from 180° to 45° and back to 180°. Dynamic biceps contractions are repeated until exhaustion, when the subject cannot continue to complete the movement. For each model, impedance was measured during the movements and the change of the impedance ΔR

between the rest muscle state (beginning of the first cycle) and exhausted muscle state (end of contractions) was calculated.

4. Results

Different states of muscle tissue are sensitive to changes in muscle electrical impedance under 50 kHz excitation [15,16]. Studies [17,18] have shown that resistance is the first parameter to change during muscle contraction, and muscle fatigue movement has limited influence on the reactance and phase. Therefore, the experiment mainly analysed the resistance parameters of the biceps muscle at 50 kHz.

4.1. Static experiment

The purpose of the static experiment was to explore the influence of the distance between the electrodes in model B electrode configuration on measured muscle electrical impedance, to verify the theoretical model B, and to find the optimal configuration of electrodes in model B with parallel electrode array.

Firstly, two measuring electrodes were placed on a biceps muscle separated by $Y = 24$ mm, as in the optimally configured Model A. The excitation electrodes were moved from the position $X = 5$ mm to $X = 20$ mm with 5 mm step. Minimal spacing between measuring and excitation electrodes $X = 5$ mm was chosen to prevent unintentional short-circuit between the electrodes during a muscle contraction. Figure 4(a) shows the relationship between the impedance measured using Imp^{TM} SFB7 and distance X between the fixed measuring electrodes pair separated by $Y = 24$ mm and the moved excitation electrodes pair. The trend of the curve in Figure 4(a) demonstrates that the muscle electrical impedance decreases as distance X increases, and the maximum impedance is measured for $X = 5$ mm distance. Therefore, we conclude that the optimal distance X between the excitation and measuring electrodes of Model B is 5 mm.

In the second experiment, influence of longitudinal distance between the electrodes was investigated, while maintaining model B optimal transversal distance $X = 5$ mm. Due to the difference in the total length of the biceps among the users and the need for miniaturization, the total longitudinal length of the electrode system is set to 90 mm. In that case, having 20 mm × 20 mm, maximal distance between the electrodes can be 45 mm. Distance Y was changed between 5 and 45 mm with a 5 mm step and measured relationship between Y and the muscle electrical impedance is presented in Figure 4(b). Figure 4(b) shows that the muscle electrical impedance increases as Y increases when X is fixed, for all volunteers. Black triangle marks represent impedance values measured using the optimally configured model A with $Y = 24$ and 10 mm ×

40 mm electrodes, and they match well with the model B $Y = 24$ mm case. Therefore, from the perspective of miniaturization and wearability, the optimal configuration of model B is using 20 mm × 20 mm electrodes and spacing $X = 5$ mm and $Y = 24$ mm, in parallel array.

In Figure 5 impedances measured during dynamic contractions for optimal model A and optimal model B are presented. It can be seen that for both models impedance consistently decreases with increased muscle fatigue. This suggests that model B can replace model A as a new electrode configuration of EIM based on the four-electrode method for muscle fatigue detection.

Between the initial state and last contraction, decrease of the measured impedance ΔR is 4–7 Ω for the optimal configuration of model A and decrease measured by the optimal configuration of model B is 8–14 Ω, depending on the volunteer, as presented in Figure 6. The experimental results show that the ΔR
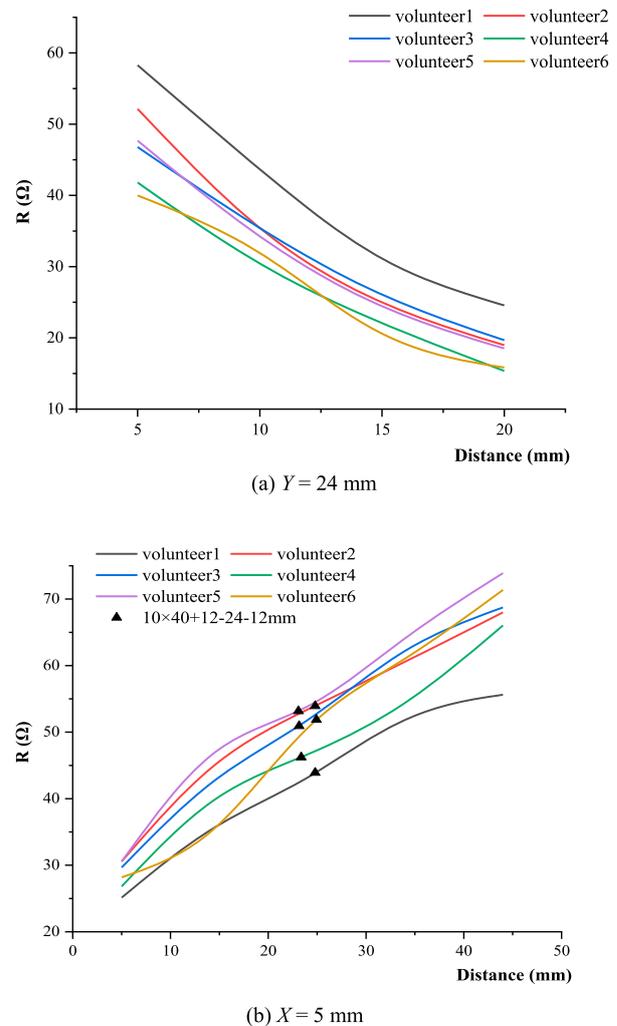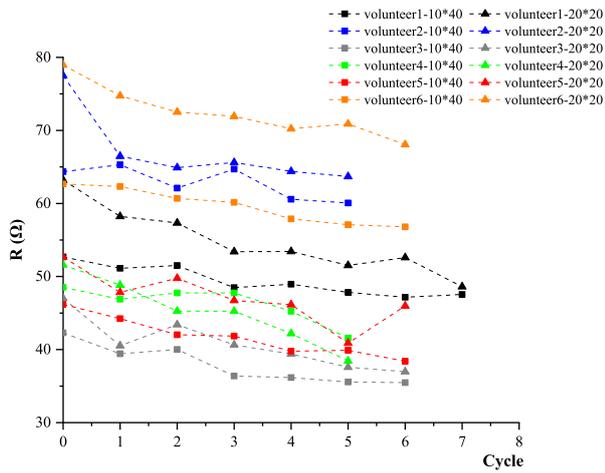

Figure 4. Measured muscle impedance on six volunteers and model B. Fixed distance between the electrodes is (a) $Y = 24$ mm, (b) $X = 5$ mm. Black triangle marks represent impedance values measured using the optimally configured model A.

Figure 5. Electrical impedance measured on biceps muscle during dynamic movements.
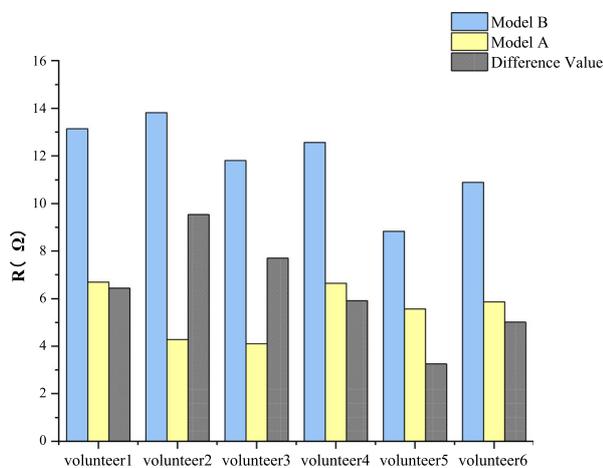

Figure 6. Decrease of muscle electrical impedance for models A and B. Difference Value is the difference of variation in muscle electrical impedance between models A and B when muscle is tired.

measured by the optimal configuration of model B is twice that measured by the optimal configuration of model A. Therefore, for the optimal configuration of EIM based on the four-electrode method for muscle fatigue detection we propose model B configuration, which features a 20 mm × 20 mm electrode size and 5–24 mm X–Y spacing, in parallel array.

5. Conclusions

There is a huge demand for wearable muscle fatigue testing in the field of sports, fitness, medical rehabilitation, and EIM muscle fatigue testing technology. In this paper, we optimized the common four-electrode configuration (model A) by changing the electrode arrangement to make the array more compact and have higher potential sensitivity (model B). An equivalent circuit model of the new four-electrode configuration has been proposed, and the *in vivo* experiments for comparing both models have been performed on the biceps muscle of six volunteers. The relationship

between the electrode configuration of model B and measured muscle electrical impedance is explored and equivalent circuit of model B is also verified. The applicability of model B to replace model A for wearable muscle fatigue detection is discussed. Change of the impedance ΔR during muscle fatigue is compared for both models. Finally, the optimal electrode configuration of EIM based on the four-electrode method for muscle fatigue detection is proposed for the development of wearable devices in the future. In future studies, the design conditions of the test equipment for wearable muscle fatigue will be considered on the basis of the proposed optimal electrode configuration. Research on the electrode material and muscle electrical impedance will be further explored.

Disclosure statement

No potential conflict of interest was reported by the authors.

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