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ELECTRICAL IMPEDANCE MYOGRAPHY FOR MUSCLE FATIGUE MONITORING

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Since 2018 the group has been working on the project “Evaluation of the Local Muscle Fatigue with the EIM Method for Wearable Applications” with the aim of developing a small wearable device for muscle fatigue monitoring.

Muscle fatigue is a phenomenon which often occurs in daily life, due to repetitive muscle flexions or long-term static contractions. It occurs when the muscle motor system cannot maintain the expected intensity required for a particular activity due to the weakening of its functional ability and is generally reversible. The continuous movement of muscles gradually reduces their work capacity, maximum contraction force, and output power [1-3]. If muscle fatigue is not properly handled, it can cause muscle strain and seriously affect the daily lives of people or, especially, the physical exercise of athletes. Muscle fatigue can be evaluated by several standard indicators, such as muscle oxygen saturation, lactic acid concentration, ultrasound image entropy, and surface electromyography (sEMG), which is most widely used [1]. However, most of these techniques are conducted in hospitals or rehabilitation centres and the patients cannot detect muscle fatigue anytime and anywhere or perform self-measurement at home. A wearable device that can monitor muscle fatigue at any time would be very useful for exercise rehabilitation, muscle disease diagnosis, sports training, and other fields.

Electrical impedance myography (EIM) is a non-invasive bioelectrical impedance technique based on the four-electrode array [4]. This technique evaluates the health status of a local muscle by applying a high-frequency, low-intensity alternating current to the muscle of interest through the outer electrodes and measuring voltage between the inner electrodes of the array, Fig. 1. EIM can also be used in clinical diagnosis and efficacy evaluation of various neuromuscular diseases. During muscle fatigue, the lactic acid content of muscle cells in muscle fibres increases, thus slowing down the conduction speed of electrical signals in muscle fibres. The EIM method detects changes in impedance due to muscle abnormalities or muscle fatigue, so it can be used for the assessment of muscle fatigue. Compared to the traditional sEMG approach for muscle fatigue assessment, the detection parameters (resistance,

reactance, phase) of EIM signals have many advantages: large EIM signal amplitude, controllable frequency, and simple pretreatment procedure. Therefore, EIM could be used as a new, low complexity and high feasibility method of real-time muscle fatigue monitoring, which can easily be integrated with various wearable devices.

Papers [1], [2] and [3] are related to the use of electrical impedance myography for the assessment of muscle fatigue on small wearable devices. In [1] the researchers described and developed a new method for estimating muscle fatigue from signals measured using four-electrode electrical impedance myography. Positions of four electrodes were optimized in [2] and a new arrangement of electrodes was proposed, optimal for estimating muscle fatigue. The new method was used to estimate muscle fatigue during static and dynamic contractions in [3]. The method for estimating muscle fatigue was modelled theoretically and by simulations and verified by *in vivo* measurements. Finally, in [5] electrical impedance myography was used for evaluating muscle fatigue induced by neuromuscular electrical stimulation.

In [1], based on the anthropometric parameters of eight volunteers, dimensions of an equivalent three-dimensional (3D) model of the entire arm were calculated. A standard four-electrode array was placed on the upper arm, above the *biceps brachii* muscle, as in Fig. 1. The model was developed in AC/DC module

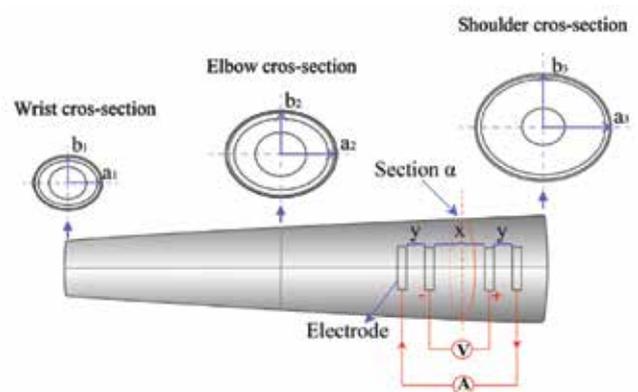


Figure 1. Human body 3D arm model has four layers (bone, muscle, fat, and skin), section α represents the cross-section at the midpoint between the excitation electrodes, [1].

of COMSOL Multiphysics 5.2a, a simulation software based on the finite element method (FEM), and used to find the optimal distance between EIM electrodes. An optimal electrode configuration allows a minimal excitation signal for a maximal measured potential difference and provides a good precondition for the front-end signal detection system of a wearable device.

A 1 mA 50 kHz current signal was used as excitation. The overall current density generated at the excitation (outer) electrodes and applied to the arm J_{arm} , current density in muscle tissue J_{muscle} , and voltage V_{sense} between the inner electrodes were calculated. From the perspective of a wearable device, a higher potential difference between the excitation electrodes leads to a higher voltage detection at the receiving end. In addition, if the signals injected into the human body can flow through the muscle layer, then the measured EIM parameters can accurately reflect the muscle characteristics. Therefore, the distance between EIM electrodes was optimized based on two parameters, which should be as high as possible: 1) the ratio of current density in the muscle layer and the current density in the whole arm (J_{muscle}/J_{arm}) and 2) the potential difference achieved at the voltage electrodes (V_{sense}). In EIM the distance between the electrodes is mainly set in three ratios of $y:x:y$ in Fig. 1: 1:1:1, 1:2:1, and 1:3:1, depending on the size of the *biceps brachii* muscles of healthy adults. In this research, the distance between the two internal EIM electrodes was set to standard $x=24$ mm, as shown in Fig. 1. The distance between the outer and internal electrodes (y in Fig. 1) was set to 8 mm, 12 mm, and 20 mm. Although according to the ratio of 1:1:1, y should be set to 24 mm, the maximum value of y was set to 20 mm because of the limited length of a volunteer's *biceps brachii* muscles.

It was shown that at a frequency of 50 kHz, the current flowing through the muscle layer accounts for more than 90% of the overall current in the arm. The value of J_{muscle}/J_{arm} increases with the space between outer electrodes, while the modulus of voltage V_{sense} decreases. The difference between the shortest (8 mm) and longest (20 mm) distances in terms of J_{muscle}/J_{arm} is less than 1 %, and the modulus of V_{sense} parameter is the largest for $y = 8$ mm. Therefore distances $y = 8$ mm and $x = 24$ mm were found optimal and used for *in vivo* EIM measurements during dynamic muscle contractions.

Before the *in vivo* measurements, the highest load a human arm can hold (maximal voluntary contraction, MVC) of eight volunteers was measured using multi-joint muscle strength assessment training system Biodex System 4, Table 1. In the *in vivo* experiments, different contraction strengths (such as 20% MVC, 40% MVC, and 60% MVC) were used to indicate different contraction states of muscles.

Volunteer	1	2	3	4	5	6	7	8
MVC, [Nm]	28.7	30.2	36.4	29.1	57.0	48.8	64.5	62.0

A block diagram and photography of a lower arm flexion experiment are shown in Figs. 2 and 3, respectively. The EIM measurement setup consisted of a signal generator (Rigol DG4162), constant current source, four electrodes placed on *biceps brachii* muscle, Agilent 1141A differential probe, and Agilent MSO7054A oscilloscope. In the experiment, the AC voltage signal with a frequency range from 1 kHz to 1 MHz and 1 V amplitude was generated and fed to the custom-built 1 mA constant current source. The current signal was then loaded into the *biceps brachii* muscle through four side-by-side electrodes of the same size (40 mm x 10 mm). The coupling voltage signal measured between the voltage electrodes at the receiving end was displayed by the oscilloscope in real-time to obtain the impedance parameters. The Agilent 1141A differential probe was used for connecting the electrodes on the body and the oscilloscope in order to solve the common ground problem between the receiving and the transmitting electrodes.

Simultaneously, real-time surface EMG signals were acquired using the Trigno Lab wireless surface EMG acquisition system. The measured sEMG signals were used to verify of EIM muscle fatigue estimation, since median frequency (MF) of power spectra of sEMG signals is often used as the parameter for muscle fatigue evaluation.

Volunteers were asked to perform repeated lower arm flexions until exhaustion while holding a dumbbell with weights of 20%, 40%, and 60% of their MVC. One dynamic contraction cycle started with the arm naturally drooping at an elbow angle of 180°. During one contraction cycle the elbow angle decreases from 180° to 45° and then increases back to 180°, as in Fig. 2. The EIM parameters were sampled every 10 cycles during this muscle fatigue process at the half-cycle point (45°).

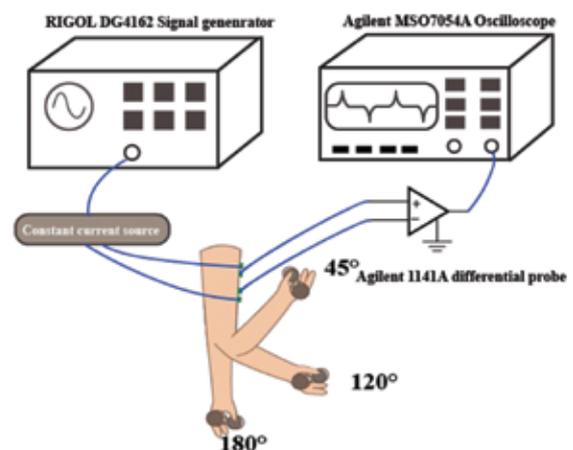


Figure 2. Block diagram of a lower arm flexion experiment, [1].

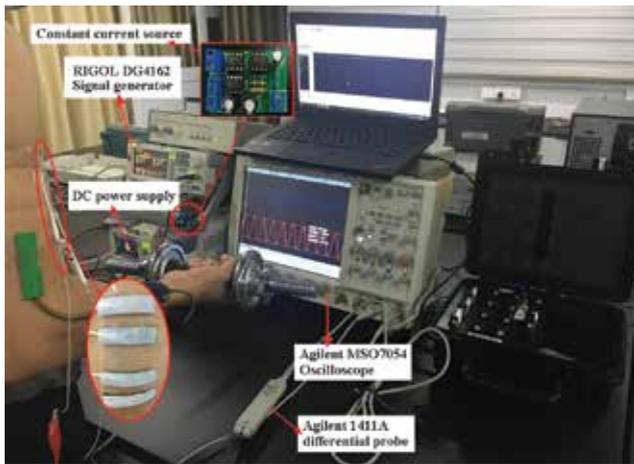


Figure 3. EIM human experiment measurement setup, [1].

In many EIM studies, resistance R is the first parameter to change during muscle contraction [6, 7] while reluctance and phase experience minimal changes. Therefore, only resistance R was used for muscle fatigue evaluation. The results were compared to median frequency of sEMG signals, which is standardly used for muscle fatigue evaluation.

Fig. 4 shows the relationship between EIM R parameters and contraction time of *biceps brachii* muscle under different loads (20%, 40%, and 60% MVC) at 50 kHz measured on the volunteer 1. The muscle R data at different load levels show a linear decreasing trend with increasing contraction time. Moreover, it is evident that the slope of the linear fit line is different for different loads. The average and standard deviation of measured resistance R on eight volunteers were calculated under different load levels and the results are shown in Fig. 5. The values of parameter R before (without fatigue) and after muscle fatigue experiment (exhaustion) were significantly different ($p < 0.01$). The difference in R before and after the experiment (drop-out value) was about 8 Ω .

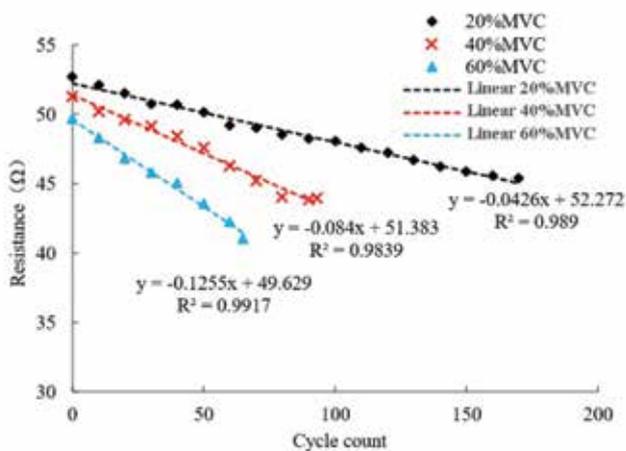


Figure 4. Variation of EIM resistance parameters with time at 50 kHz for three different contraction forces, [1].

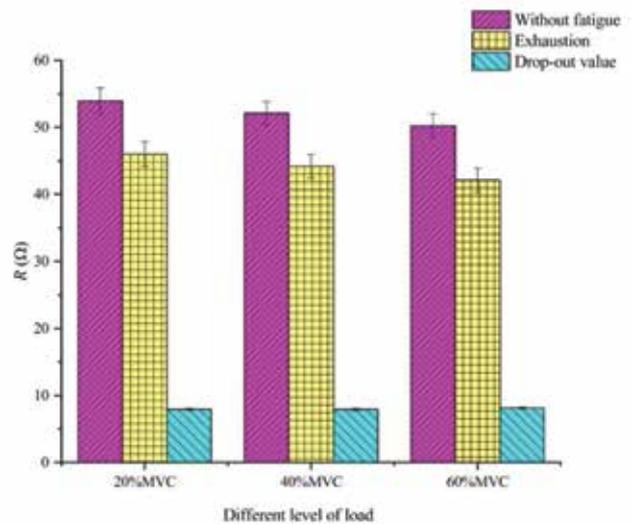


Figure 5. Average value and standard deviation of EIM resistance measured on eight volunteers during fatiguing dynamic contractions, [1].

The measured sEMG median frequency (MF) for different loads also follows the same downward trend as EIM R parameter, i.e. its value decreases as fatigue level increases, and a higher load results in a more rapid decline.

EIM and sEMG signals were acquired simultaneously during dynamic contractions until exhaustion. The achieved number of contractions cycles decreased with increasing load. Muscle fatigue index (MF) calculated from sEMG signals (blue) and R measured using EIM electrodes (red) for 40% MVC load normalized to the number of cycles are shown in Fig. 6. It has been experimentally shown that the absolute values of the linear fit slope are higher for the heavier load and the decline slope of R linear fit line for each load is nearly 2/5 that of the MF, [1]. Thus, the decreasing rate of the R is not as fast as a decrease in MF, but a considerable

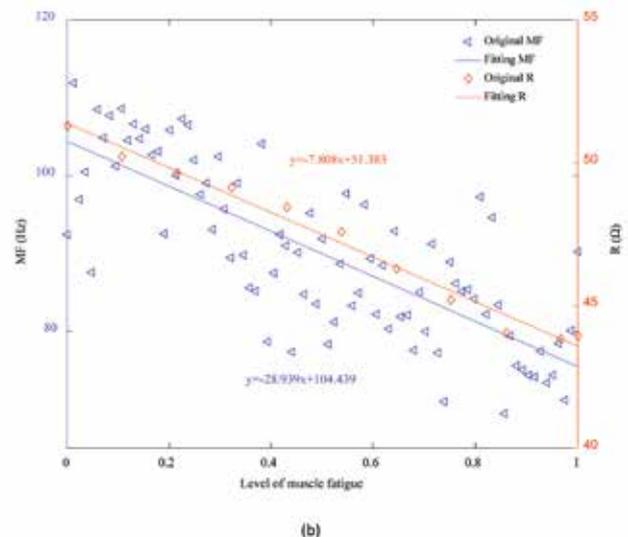


Figure 6. Median frequency (MF, blue) and impedance R (red) measured during dynamic contractions for 40% MVC, [1].

consistency is observed in the decline regularity of the two curves. Therefore, EIM with a standard electrode four-in-line arrangement can be used for muscle fatigue evaluation, and the rationality of this approach was verified. In the muscle dynamic contraction fatigue experiment, the subjects were considered to reach the semi-fatigue point when the measured R parameters decreased by approximately 4Ω . With the further R decrease approaching 8Ω , the muscle fatigue is considered to reach its limit. Thus, following the magnitude of R decline the subjects can adjust their exercise intensity and avoid muscle fatigue or the damage caused by excessive exercise.

After proving the feasibility of exploiting EIM for muscle fatigue evaluation, the researchers further investigated and optimized influential parameters of EIM (electrodes arrangement and distance) based on equivalent circuit [2] and finite element [3] models.

The standard four-in-line EIM electrode arrangement has low potential sensitivity and occupies a large area, which is not suitable for the miniaturization and portable design of wearable devices, so in [2] a new more compact electrode arrangement was proposed, Fig. 7. The surface of electrodes in both models is the same, but their shape is different. In model A, standard $40 \text{ mm} \times 10 \text{ mm}$ rectangular electrodes are used arranged in a serial array. In model B $20 \text{ mm} \times 20 \text{ mm}$ square electrodes are used, arranged in a parallel array. An equivalent circuit model of both models was proposed and the influence of distances between the electrodes of model B was investigated. The results were tested *in vivo* on six volunteers using ImpTM SFB7, a device for measuring the response voltage while generating a 2 mA current signal.

Two sets of static experiments were performed on six volunteers. In the first set, measuring electrodes were

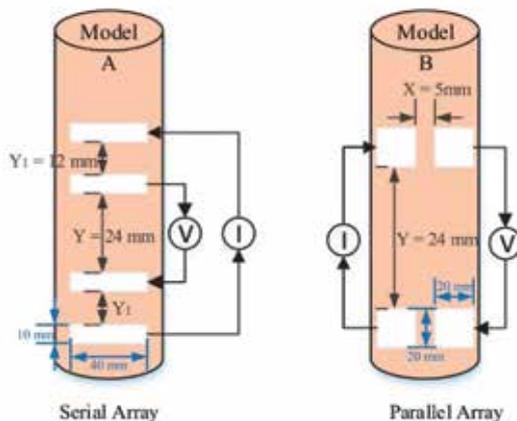


Figure 7. EIM electrode configurations based on the four electrodes. Model A is the traditional four-electrode configuration with a serial array. Model B is a proposed four-electrode configuration with a parallel array, [2].

placed on a biceps muscle separated by $Y = 24 \text{ mm}$, and excitation electrodes were moved from the position $X = 5 \text{ mm}$ to $X = 20 \text{ mm}$ with a 5 mm step. The maximum impedance was measured for $X = 5 \text{ mm}$, so in the second 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. In this experiment the muscle electrical impedance increased as Y increased when X was fixed, for all volunteers. Therefore, from the perspective of miniaturization and wearability, the optimal configuration of model B presented in Fig. 7 right is using $20 \text{ mm} \times 20 \text{ mm}$ electrodes and spacings $X = 5 \text{ mm}$ and $Y = 24 \text{ mm}$, in parallel array.

The second set of *in vivo* experiments involved measurements of muscle impedance measured using optimal model A and model B configurations in Fig. 7 during dynamic contractions (lower arm flexion, as in Fig. 2). Dynamic biceps contractions were repeated until exhaustion, when the subject cannot 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. For both models impedance consistently decreased with time, i.e. with increased muscle fatigue. Between the initial state and the last contraction, the decrease of the measured impedance ΔR was $4\text{--}7 \Omega$ for the optimal configuration of model A and the decrease measured by the optimal configuration of model B was $8\text{--}14 \Omega$, depending on the volunteer, as presented in Fig. 8. The experimental results show that the ΔR measured by the optimal configuration of model B is twice that measured by the optimal configuration of model A, so model B configuration can be used for EIM-based muscle fatigue detection

Further investigations on the optimal model B (parallel) configuration were performed in [3], by means of

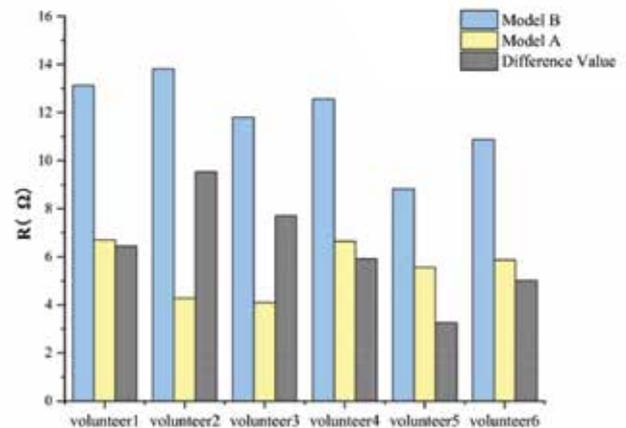


Figure 8. Decrease of muscle electrical impedance for models A and B. Difference Value is the difference of variation in muscle electrical impedance between models A and B when muscle is tired, [2].

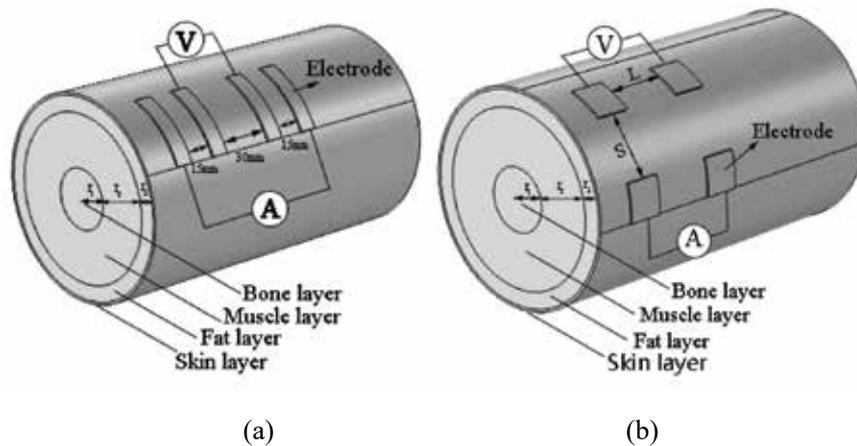


Figure 9. A four-layer FEM model of human upper arm: (a) serial electrode configuration method; (b) parallel electrode configuration method, [3].

numerical simulations in COMSOL and *in vivo* measurements. A four-layer FEM model consisted of skin, fat, muscle, and bone layers, Fig. 9, and was based on anthropometric characteristics of 10 volunteers. Comparing serial and parallel configurations with the equal electrode area and including approximately the same muscle area, it was again proven that parallel configuration results in a larger EIM impedance amplitude at all tested frequencies (10, 25, 50, and 100 kHz). The larger impedance implies a higher detection amplitude, which is easier to integrate into wearable devices.

Apparent impedance is the electrical impedance measured on the skin surface. Its value depends on the contribution of each layer below the electrodes, which depends on the electrode characteristics (size, material, shape, and position on the skin). Sensitivity analysis was used to calculate the impedance of each layer and optimize positions of the four electrodes on the skin surface (distances S and L in Fig. 9) in order to maximize

the percentage of muscle tissue contribution to the apparent impedance. It was shown that by adjusting the electrode spacing on the skin surface, the contribution of the muscle tissue in the apparent impedance changes. At 50 kHz, when $S = 5$ mm and $L = 5$ mm, the contribution of the fat layer in apparent impedance was 81.05%, and the muscle layer was 19.18%. However, for larger electrode spacing $S = 20$ mm and $L = 20$ mm, the contribution of the fat layer decreased to 28.67%, and the contribution of the muscle layer increased to 71.68%. For the bone and skin layers, the difference was small and changed only within 1%.

In order to reduce the influence of individual fat differences on EIM results, the rate of change of EIM impedance with fat thickness was studied for different electrode distances. Fat thickness was set to 6 mm (base value, 1.0x) and thickened with a step size of 3 mm. The results for the fat thickness of 6 mm (1.0x), 9 mm (1.5x) and 18 mm (3.0x) are shown in Fig. 10. The selectivity of impedance to muscle tissue Sel_m

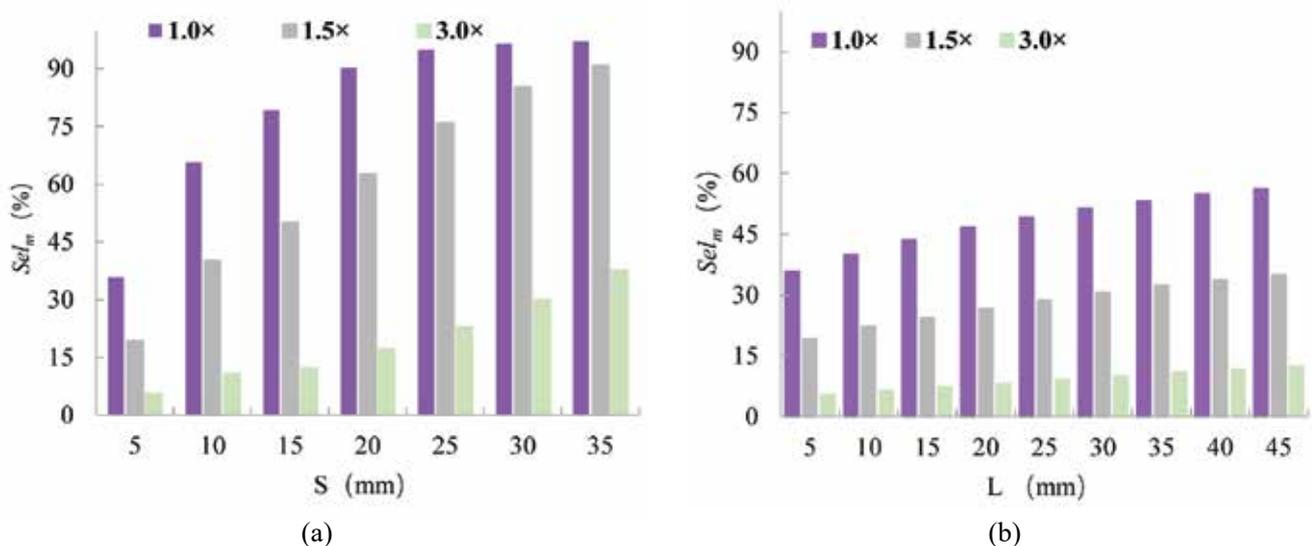
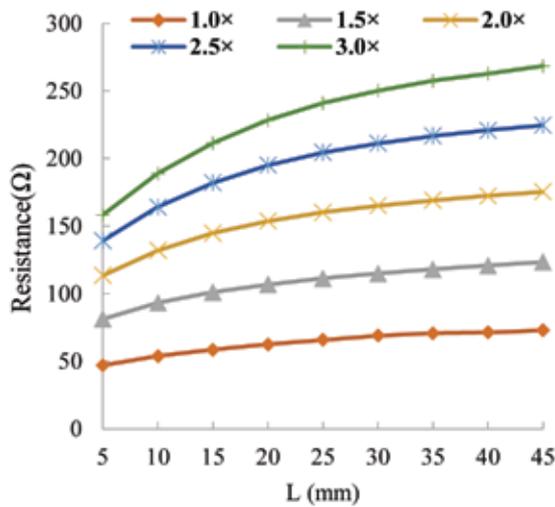


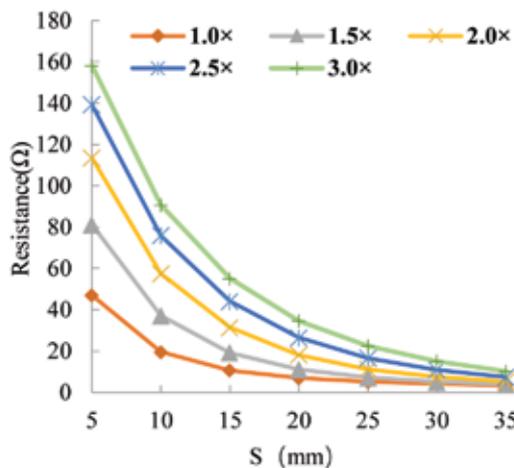
Figure 10. (a) The impedance selectivity of EIM variation with S . (b) The impedance selectivity of EIM variation with L , [3].

increased with increasing S and L and was not influenced by fat thickness. Furthermore, Sel_m increased much faster with increasing S compared to L . Taking the fat thickness of 1.5x as an example, for spacings $L = 5$ mm and $S = 35$ mm, Sel_m was 90.95%, while at $S = 5$ mm and $L = 35$ mm, Sel_m was 32.34%.

The variation of EIM resistance with S and L and different fat thicknesses (between 6 mm, 1.0x and 18 mm, 3.0x) is shown in Fig. 11.



(a)



(b)

Figure 11. (a) The resistance stability of EIM variation with L . (b) The resistance stability of EIM variation with S , [3].

It was shown that at $L = 5$ mm and $S = 5$ mm (Fig. 11 a), the increase of resistance was 236.6% when the fat thickness increased from 1.0x to 3.0x. The increase of resistance was 423.3% with increasing fat thickness from 1.0x to 3.0x when $L = 45$ mm. The resistance increased with increasing L and fat thickness, indicating that the resistance stability decreased with increasing L . At $S = 5$ mm and $L = 5$ mm (Fig. 11 b), the resistance increased 236.66% as the fat thickness increased from

1.0x to 3.0x. The increase of resistance was 418.91% with increasing fat thickness from 1.0x to 3.0x when $S = 15$ mm and $L = 5$ mm. Although the resistance change showed a decreasing trend with increasing S , the resistance value decreased rapidly for smaller S and slowly for larger values of S .

Considering requirements of the EIM electrode design for the parallel electrode configuration, such as the overall area of the electrodes, impedance amplitude, stability, and selectivity Sel_m (muscle tissue contribution), distances $S = 10$ mm and $L = 20$ mm were finally chosen for the *in vivo* experimental study of local muscle fatigue. The EIM experimental platform is presented in Fig. 12. The EIM electrodes were placed in the middle of a *biceps brachii* muscle and connected to the Imp™ SFB7 anthropometric analyser working in the impedance spectrum mode. The angle between the upper and lower arm was constantly monitored using an angle sensor and the EMGworks software, in order to avoid changes in muscle length caused by inadvertent angle change during the experiments. Volunteers held dumbbells of different weights (0.1, 0.3, and 0.5 of their MVC) in their right arm. In the static experiment the angle between the lower arm and the body was $\theta = 45^\circ$, while in the dynamic experiment it changed between $\theta = 0^\circ$ and $\theta = 135^\circ$, Fig. 13.



Figure 12. EIM experimental platform, [3].

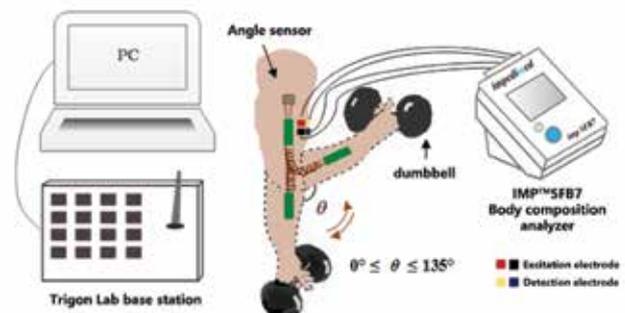


Figure 13. Block diagram of dynamic experiment, [3].

The trend of change in resistance during static contractions is shown in Fig. 8. T_s was the time needed

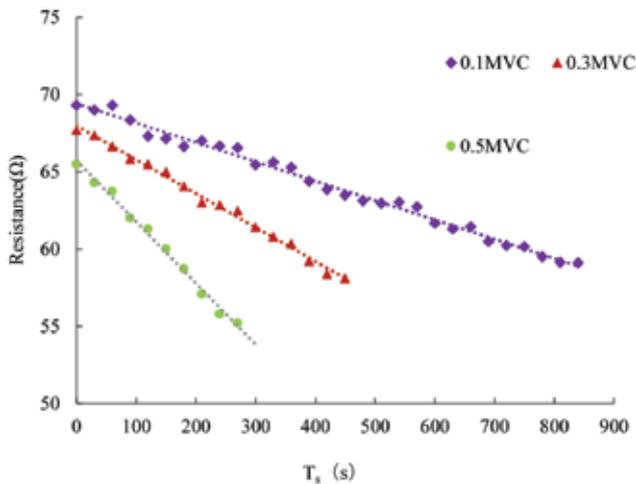


Figure 14. Static contraction experiments of muscles under different load conditions (0.1, 0.3, and 0.5 MVC). The decrease of R (50 kHz) measured on Volunteer #1 from complete relaxation to extreme fatigue, [3].

for the muscle to change from a resting state to a fully fatigued state. When the muscle was in a state of complete fatigue, the EIM impedance remained constant. At $f = 50$ kHz the EIM impedance value had a linear correlation with the degree of muscle fatigue, while the resistance had a negative correlation with the degree of fatigue, as in [1]. In addition, fatigue accumulation occurred more rapidly at higher MVCs, resulting in a higher slope of the linear curve and a shorter duration of the experiment T_s . For example, $T_s = 840$ s under 0.1 MVC load, and $T_s = 270$ s under 0.5 MVC load. To avoid individual differences, static experiments were performed on 10 volunteers. Although the initial EIM impedance values showed large differences, since they depend on the fat level of the volunteers, the resistance reduction between relaxed and exhausted muscle was about 10Ω .

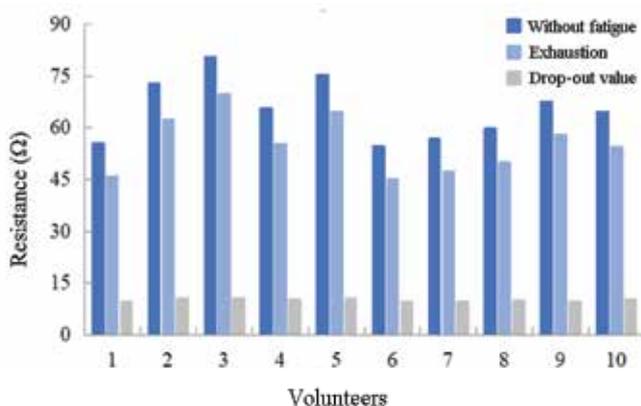


Figure 15. Resistance changes of 10 volunteers before (without fatigue) and after (exhaustion) the dynamic experiment for 0.3 MVC load, [3].

During dynamic contraction experiments the measured electrical impedance had a linear relationship with the number of contraction cycles, i.e. the duration of

the experiment, for all three loads, which is consistent with the static experiment results. For heavier loads the experiments lasted shorter, indicating that the volunteer's fatigue progress was faster. The *biceps brachii* reached a completely fatigued state when the change in the amplitude of the EIM impedance value between two contraction cycles was extremely small and almost constant. EIM resistance R measured on ten volunteers before and after the dynamic contraction experiments for 0.3 MVC load are presented in Fig. 15. A change of resistance between the beginning and the end of the exercise (exhaustion) varied around 10Ω for all volunteers.

One of the first applications of a newly developed muscle fatigue monitoring technique was published in [5]. Neuromuscular electrical stimulation (NMES) is a common method for rehabilitation treatment and sports training, in which low-amplitude current is applied to the muscle tissue. However, a common side effect of the stimulation is muscle fatigue. The research in [5] focused on the changes in EIM parameters (impedance amplitude $|Z|$ and phase θ) under muscle fatigue induced by the NMES. The effects of different stimulation parameters (amplitude, frequency, and pulse width) on NMES muscle fatigue were evaluated and analysed. The EIM parameters were compared to the mean power frequency (MPF) of sEMG thus verifying the validity of using EIM for evaluation of the muscle fatigue induced by NMES.

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