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Advancements and Challenges in Electrical Impedance Myography (EIM): A Comprehensive Overview of Technology Development, Applications in Sports Health, and Future Directions

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ABSTRACT ElectricalImpedance Myography (EIM) is an innovative, non-invasive technique offering a convenient means of localized exogenous electrophysiological recording. By measuring muscle impedance parameters, this method characterizes the physiological state of muscles, functioning as a biomarker for muscle contractility, injuries, and the progression of neuromuscular diseases. This paper provides an overview of the current state of EIM technology development, along with modeling and data analysis methods, focusing on their application requirements. It further highlights the advancements in EIM research within the realm of sports health, emphasizing its efficacy in identifying injuries and monitoring wound healing, and discusses existing technological limitations. Additionally, the paper explores future research directions. Serving as a transient biosensor during physical activity, EIM holds significant potential in sports health. It presents a promising alternative to invasive and costly clinical assessment methods, positioning itself as a viable personal monitoring tool for both professional athletes and fitness enthusiasts. Nevertheless, the resolution of technical challenges and the establishment of industry-standard implementation programs are essential prerequisites for EIM to evolve into a standard clinical assessment tool.

INDEX TERMS Clinical assessment tool, electrical impedance myography, electrophysiological recording, muscle physiological state, sports health.

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I. INTRODUCTION

In sports activities, prolonged and inappropriate exercise can easily lead to muscle fatigue, which can adversely affect athletes' performance and muscle contraction capabilities [1]. A proper exercise program is crucial for fitness enthusiasts who want to improve their physical fitness and overall health. However, prolonged muscle fatigue and excessive exercise can disrupt the delicate balance between the body's oxidative and antioxidant capacities, which can lead to oxidative stress damage and deterioration of bodily functions [2]. Excessive muscle training carries the risk of chronic fatigue, musculoskeletal disorders, and even disability [3], [4], directly affecting athletes' careers and physical well-being. Moreover, rapid injury diagnosis in the sports field is crucial to guide physicians to timely interventions and thus avoid treatment delays. Rapid determination of injury recovery is closely linked to the athlete's return to play (RTP) and plays a pivotal role in reducing the incidence of secondary injuries [5]. Consequently, there is an urgent need for objective, quantitative, and easily accessible assessment tools for evaluating muscle fatigue and muscle status in the field of sports health.

One promising technology that addresses these needs is electrical impedance myography (EIM), an innovative exogenous electrophysiological detection technique. EIM is applied in the clinical diagnosis and efficacy evaluation of various neuromuscular diseases [6]. Introduced by Professor Seward B. Rutkove in 2002 [7], EIM analyzes local electrical impedance to diagnose neuromuscular diseases. Early studies by Shiffman et al. in 2003 [8] demonstrated significant quantitative and qualitative differences in impedance parameters between patients with neuromuscular diseases and a healthy group, validating the clinical potential of EIM. Unlike methods that rely on the intrinsic bioelectrical activity of muscles, EIM depends primarily on the changes in the electrical properties of the internal structure and tissue layers of the muscle in response to applied current signals at different measurement frequencies and directions [9], [10]. To date, EIM has been used primarily as a biomarker for neuromuscular diseases, including the identification of disease severity and assessment of efficacy following therapeutic interventions that have progressed over time. Several studies have shown that the muscles of patients with amyotrophic lateral sclerosis [11], [12], radiculopathy [13], Duchenne muscular dystrophy [14], [15], muscle disuse atrophy [16], and stroke [17] are significantly altered from healthy muscles, exhibiting variability in resistance, reactance, and phase. This is mainly due to significant changes in the conductivity and dielectric constant of the diseased muscles [9]. At the same time, changes in tissue volume, including muscle fiber atrophy and edema, also contribute to the partial changes in EIM [18]. Consequently, EIM performs an important effect in the clinical assessment of muscle function and muscle physiologic status. However, the focus of this paper is not on the applicability of EIM in neuromuscular diseases, as reviews on this topic can be referred to [19], [20].

As a rapid, non-invasive muscle assessment tool, EIM has also been used for muscle health quality assessment and adaptive fitness studies in sports and exercise. The main objective of this review is to summarise EIM's technological developments and research directions applicable to the field of sports health. A prevalent perspective suggests that currents in the α-dispersion frequency range travel primarily through the extracellular fluid [21], making the resistance (R) in this band indicative of tissue dehydration or edema status. In the β-dispersion range, high-frequency currents can flow through the cell's interior, providing information about cellular health and integrity [22]. Given these insights, bioimpedance techniques, frequently employed in sports, have been utilized to analyze athletes' hydration, whole-body or segmental body composition, and muscle mass [13], [23], [24], [25]. EIM has recently gained traction in sports activities and physical fitness, serving to analyze muscle contraction, identify muscle fatigue, and classify injuries during training. Meanwhile, a number of studies have shown that EIM is also worth exploring in sports health, as it is a fast and accurate muscle contraction sensor that captures the instantaneous changes occurring in muscle contraction [8], [26], [27], [28]. For example, a rapid decrease in impedance parameters during training with loads may indicate the onset of muscle fatigue, which can serve as an early warning sign of muscle damage [18], [29], [30], [31]. EIM is an economical, convenient, and accurate assessment tool for injury detection and recovery follow-up that can be used to predict RTP for athletes [32], [33], [34], [35], [36]. Its potential applications in sports health are foreseeable, with advantages including high signal amplitude, controllable frequency, ease of signal acquisition and data processing. In addition, its non-invasive measurement method makes it very suitable and affordable for integration into wearable devices, promising accurate muscle assessment for professional athletes and rapid muscle evaluations for the general public in daily life.

While EIM has been widely explored in clinical disease assessment, sports injury assessment, fatigue monitoring, sports rehabilitation, and bionic control, it is poised to evolve into an objective and standardized technique for assessing muscular status in the future. Despite the myriad advantages of EIM, the existing studies exhibit a wide variation in procedures due to the unstandardized nature of the technique and other reasons. Consequently, EIM is yet to be standardized as a muscular status assessment tool in both clinical and the industrial settings. This review aims to elucidate the developmental status and influencing factors of EIM in the sports health domain, outlining research procedures employed for assessing muscular status in athletic training, presenting existing research findings, and proposing future research directions. The structure of this paper is shown in Fig. 1.

II. EIM TECHNOLOGY OVERVIEW

A. IMPLEMENTATION PRINCIPLE OF EIM

At a microscopic level, the cell is the basic structural and functional unit constituting an organism [37]. Cells are infiltrated in the interstitium; the cytoplasm and interstitium are predominantly fluid substances, often referred to as intra- and

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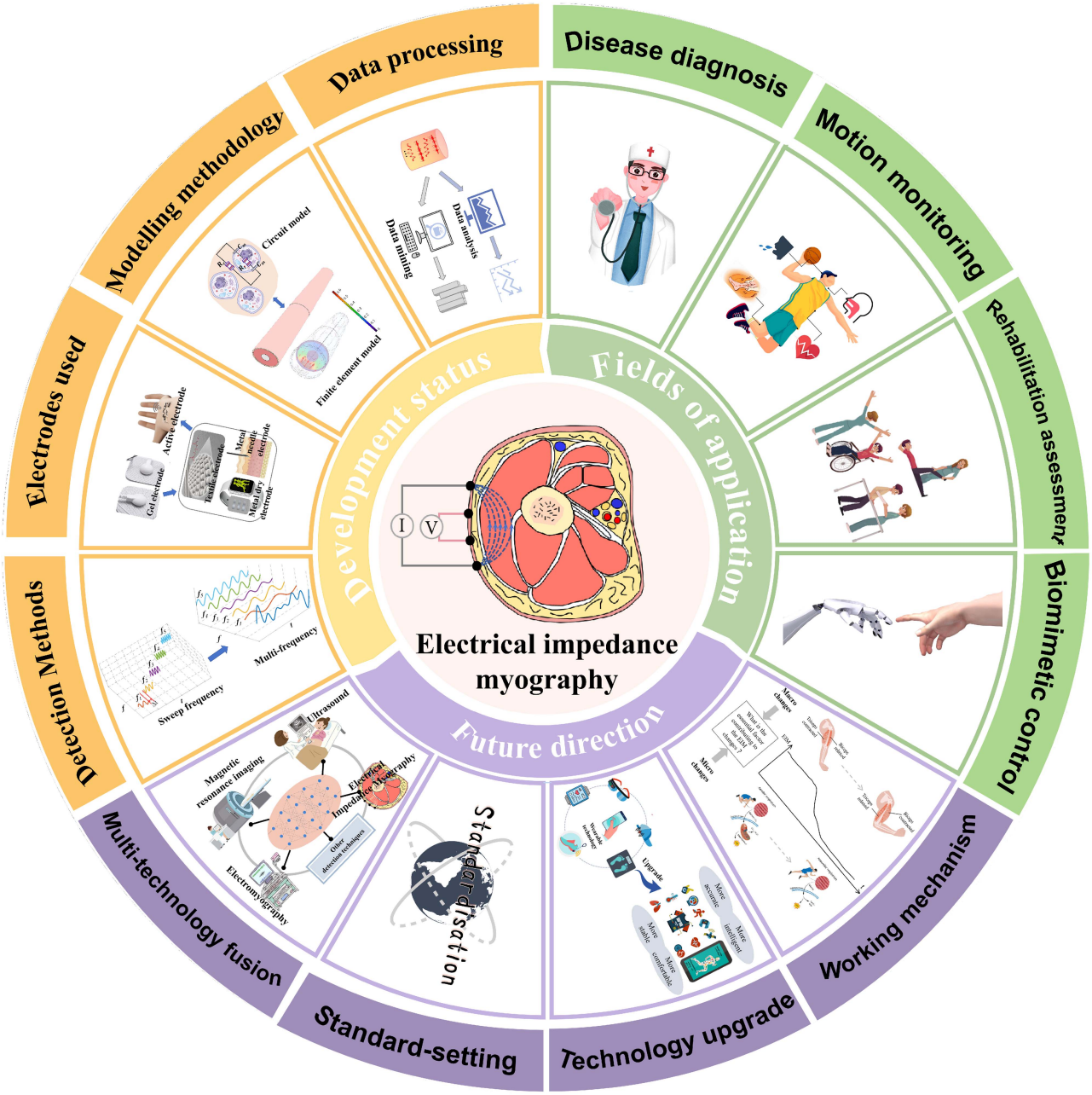

FIGURE 1. **Structure of the paper. This paper focuses on the current development status, application areas and future development direction of the EIM technique.**

extracellular fluids, which are electrolytes with good electrical conductivity [38]. The cell membrane is an elastic and selective semi-permeable membrane composed of phospholipids, which is mainly responsible for exchanging substances in the cytoplasm and interstitium to achieve nutrient uptake and metabolic waste elimination, and with this membrane exchange property maintains the body's electrical conductivity [37], [39]. The electrical properties of tissues mainly include electrical conductivity σ, magnetic permeability μ and relative capacitance (or permittivity) ε, where the magnetic permeability is negligible from 10^{-7} to 10^{-5} H/m. Schwan et al. [40] elaborated on the changes in the electrical behaviour of tissues in specific frequency intervals and proposed the frequency dispersion theory. The theory classifies the frequency response of human tissues to external sources into four different regions α, β, δ and γ, the electrical properties of the main types of tissues areas shown in Fig. 2. The α-dispersion (1 kHz–100 kHz) is mainly concerned with ion migration in the extracellular fluid, which is usually associated with diffusive changes in the ionic layer of the cell membrane in non-homogeneous tissues, where the cell membrane is a more significant impediment to the current. In this frequency range, the conductivity of the tissue changes more steadily, while the relative capacitance shows a clear tendency to decrease. The β-dispersion is in the RF band (100 kHz–10 MHz), where ion channel interactions begin to emerge, and

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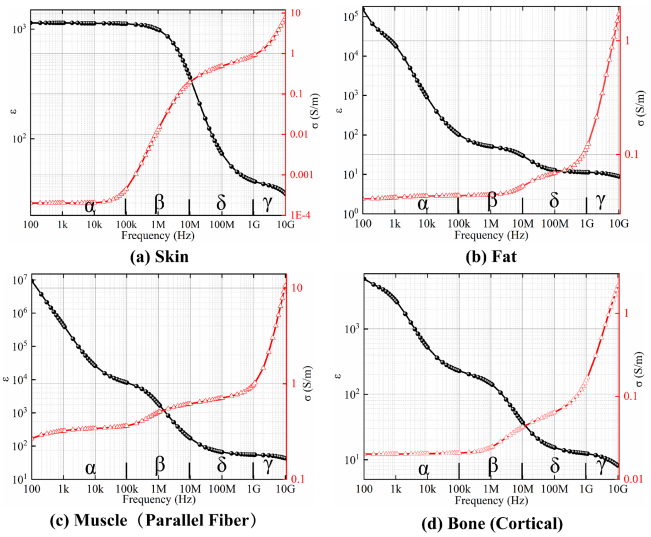

FIGURE 2. Electrical conductivity and relative capacitance of skin, fat, muscle and bone, with horizontal and vertical coordinates taken in logarithmic coordinates. Tissue conductivity is proportional to frequency and relative capacitance is inversely proportional to frequency. (a) Skin; (b) Fat; (c) Muscle (Parallel Fiber); (d) Bone (Cortical).

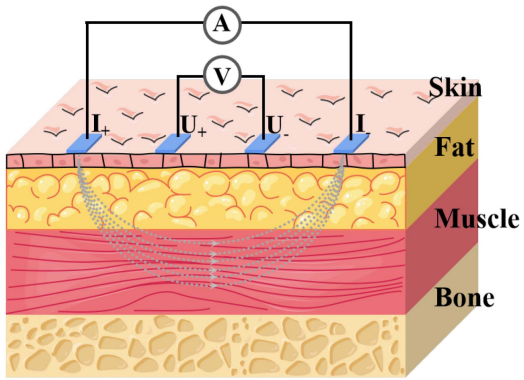

FIGURE 3. Principle of EIM. The outer two electrodes are injected with alternating current, which will flow through the various layers of tissue of the target muscle, and due to the difference in conductivity, the current will flow predominantly through the muscle layer, with the voltage response ultimately obtained from the inner two electrodes.

the capacitance effect of the cell membrane becomes essential, allowing more current to pass through the cell membrane. In the δ-dispersion (10 MHz–1 GHz), small molecules and proteins' fast polarisation and relaxation processes inside the cell membrane play a dominant role. Finally, in γ-dispersion (above 1 GHz), the electric dipole moment of free water becomes a significant factor, and the polarisation and relaxation processes of water molecules inside and outside the cell dominate. Although this review focuses on the bioimpedance technique in the α and β bands, the application of the technique at higher frequencies will also be briefly discussed to enlighten the reader on the application of the technique in the microwave band. In δ and γ dispersion, water plays a decisive role in the electrical properties of tissues, biological small molecules and proteins. Therefore, the bioimpedance technique in the microwave band (δ and γ) is sensitive to different properties of tissue stratification. The study [41] showed that it is possible to discriminate the percentage of tissue components, including adipose, glandular, and fibroconnective, using permittivity at 5 GHz. Bioimpedance techniques in the microwave band from 200 MHz to 13.6 GHz could discriminate breast cancer cells [42], and bioimpedance at 50 MHz to 900 MHz showed the highest discrimination between malignant and normal tissues in the breast [43]. In addition, blood has a high water content, so the bioimpedance technique has also been used in the high-frequency band to monitor blood changes [44]. In addition, microwave band dispersion can ignore the effect of cell membranes in probing the intracellular status, and it is also used in industry to detect the survival of bacteria [45].

α and β dispersion are mainly changes in electrical properties caused by ion diffusion at the ends of the cell membrane and by the membrane's capacitive effect. Most current research has focused on these frequency bands, as the impedance in this band reflects essential information about the structure and composition of the tissue itself and its relaxation behaviour with time [46]. EIM is a bioimpedance measurement technique located in the α and β bands for electrophysiological muscle detection [6]. It utilizes the differences in electrical properties between biological tissues to obtain and analyze physiological signals from the organism. This technique involves the injection of low-energy alternating current excitation signals within the frequency range of kHz to MHz into specific muscles or muscle groups of interest. This is accomplished through two outer excitation electrodes positioned on the skin surface above the muscle of interest, as shown in Fig. 3. As each biological tissue has unique electrical characteristics, a voltage is generated in the tissue as a response to the injected current. The two inner electrodes on the skin surface above the muscle of interest measure the generated voltage, enabling the calculation of the tissue electrical impedance value. This process facilitates the analysis of both physiological and pathological characteristics inherent to the targeted biological tissue. Not using the entire α frequency band is because tissues are more susceptible to damage at low-frequency excitation. For example, the risk of ventricular fibrillation is highest at frequencies of 10–200 Hz, but the risk decreases rapidly at frequencies above 1 kHz [47]. In modern biomedical engineering, the application of bioelectrical impedance detection has become a standard method for extracting physiological and pathological information from biological tissues. Renowned for its harmlessness, non-invasiveness, cost-effectiveness, ease of operation, and wealth of functional information, this technique has proven to be invaluable. In the following section, the current development and future research directions of EIM are presented.

B. DETECTION METHODS

The bridge method employing a balanced circuit, as illustrated in Fig. 4(a), represents an early approach to measuring bioimpedance. The impedances of the four legs of the bridge are Z_1, Z_2, Z_3, and Z_x, respectively, where Z_x is the impedance to be measured. In addition, S is the excitation source of the

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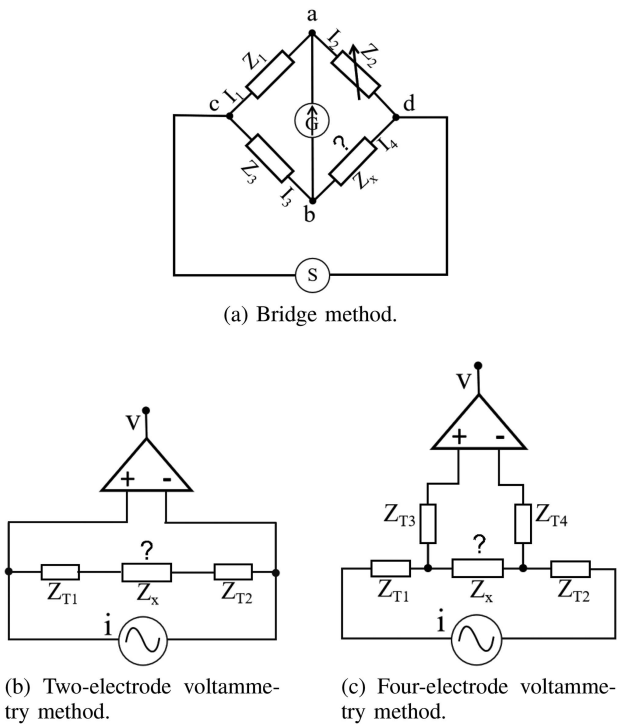

(a) Bridge method.

(b) Two-electrode voltammetry method.

(c) Four-electrode voltammetry method.

FIGURE 4. EIM detection methods.

bridge and G is the bridge balance indicator. The bridge is considered balanced when the bridge balance indicator reads 0, indicating equal voltage between points a and b. From the calculation of Kirchhoff's law, we can find that $I_1 = I_2$, $I_3 = I_4$, and therefore $Z_1/Z_2 = Z_3/Z_x$. Although effective, this method's complexity in achieving equilibrium and its inherent inaccuracies have limited its widespread use [48].

Voltammetry, a widely used impedance measurement method, injects a current signal into tissue and measures the resultant voltage, as in Fig. 4. In the two-electrode voltammetry method, the same pair of electrodes are used to inject a current signal into the tissue and measure the response voltage, as shown in Fig. 4(b). However, this method is susceptible to polarization phenomena since the current density is the highest around the electrodes, and also the contact impedance between the skin and the electrodes is high, resulting in a significant measurement error [49]. Nowadays, the four-electrode voltammetry method is commonly used as a substitute, which means that the injection of the current signal and the measurement of the response signal are performed by two pairs of electrodes, as shown in Fig. 4(c). In this electrode arrangement the current distribution is more uniform, and there is no polarization phenomenon of the electrodes. In addition, the interference signal generated by the contact impedance is much lower than that of the biological tissue, which improves the measurement accuracy.

Traditionally, EIM has been measured using a four-electrode impedance method at a fixed frequency of 50 kHz. In 2002, Rutkove et al. [7] assessed the muscular status of normal subjects compared to patients with different neuromuscular disorders based on a localized bioimpedance technique at a frequency of 50 kHz. They also provided a reference EIM value for five muscles in 50 healthy subjects at a frequency of 50 kHz [50]. This frequency was chosen because skeletal muscle tends to be most reactive at 50 kHz and most affordable impedance measurement devices provide current at this specific frequency [51]. However, the electrical properties of biological tissue vary with frequency and direction, and the amount of information obtained from single-frequency analysis is relatively limited, so scan-based EIM measurements are an alternative. It sends excitation signals at different frequencies at intervals over some time, sacrificing the time domain to obtain muscle impedance information at multiple discrete frequencies. Currently, most commercial broadband bioimpedance devices sold on the market are based on the implementation of frequency scanning, such as the Keysight E4990 A impedance analyzer in the United States [52] and the SFB7 introduced by ImpediMed in Australia, which can sample impedance values in the frequency range from 3 kHz to 1000 kHz [53]. Although this method provides additional information in the frequency domain, due to certain time intervals between frequency points for transmitting, it fails to capture continuous and rapidly changing dynamic processes. Neither single-frequency nor scanned-frequency measurements effectively address the compatibility of frequency-domain and time-domain information, which means that they each have limitations in their ability to demonstrate the frequency characteristics of signals concerning changes over time. As a result, both methods are somewhat limited in their applicability in scenarios requiring real-time motion monitoring or other rapid dynamic changes.

Recognizing the need for capturing the time-varying information in dynamic systems, multi-frequency EIM has emerged as a mainstream muscle impedance measurement method. This method allows the measurement of impedance information at multiple frequency points within a signal cycle, which allows for a more comprehensive understanding of muscle impedance dynamics. Bragos et al. [54] designed a dynamic myocardial state measurement system with an acquisition time of less than 2 ms based on a multi-sinusoidal impedance spectrum, which represents the first attempt to characterize myocardial impedance during the cardiac cycle. Min [55] and Sanchez [56] discussed the methodology of designing excitation signals based on a short linear frequency modulation pulse and a multi-sinusoidal signal with a broadband spectrum, respectively. In 2021 our team proposed a parallel impedance measurement method based on the LabView platform for measuring the electrical characteristics of humans [57]. First, we used the fast Fourier inverse variation to generate a multi-frequency excitation signal by superimposing multiple frequency signals from 10 kHz to 1 MHz in the time domain. Then the signal measurement unit recovers the individual frequency components by fast Fourier transformation after passing through the tissue to be measured. The system measures in 0.1 ms, which is determined by the minimum excitation frequency, and is realized as shown in Fig. 5(b) [57]. This method is mainly based on the

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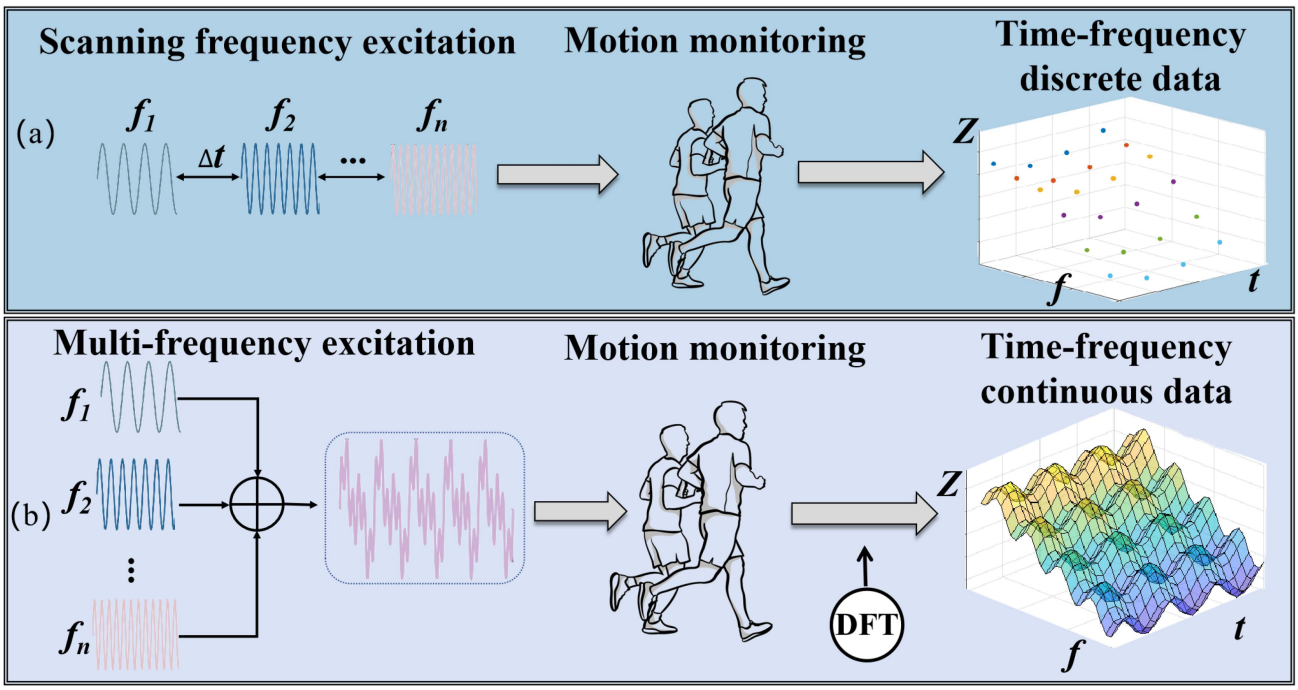

FIGURE 5. (a) Scanning frequency measurement; (b) Multi-frequency measurement. Scanning frequency measurement involves sending out excitation signals with different frequencies at intervals to obtain discrete time-frequency information. On the other hand, multi-frequency measurement involves sending out multiple frequency excitation signals simultaneously, and the signal receiving end obtains continuous time-frequency information after processing the mixed signals using Discrete Fourier Transform (DFT).

principle of orthogonal frequency division multiplexing (OFDM). Excitation signals with different frequencies are sent simultaneously in one signal period so that the signal with the lowest frequency determines the measurement time. The method utilizes broadband resources to improve time efficiency, so with the proper design of the mixed signals, this method can capture the dynamic changes of muscle impedance. The technique requires a lot of computation on the sampled signals at the receiver side, so improving the computational efficiency is a crucial issue to be addressed in the future. If the technique is to be used in a lightweight wearable system, reducing power consumption is another priority, taking into account the comfort of the user and standby time.

C. MEASURING ELECTRODES

Electrodes are an essential part of signal acquisition in electrophysiological sensing research. This section describes the types of electrodes used in bioimpedance measurements and the requirements of electrodes for wearable and dynamic physiological monitoring.

Typically, electrodes used for physiological signal acquisition can be simply categorised into two types: dry electrodes and wet electrodes. Wet electrodes require conductive media such as electrolytic gels and conductive pastes to form a conductive path between the skin and the electrode. However, the electrode materials of dry electrodes are usually metals or semiconductors that are in direct contact with the skin without the need for adhesives [58]. Disposable Ag/AgCl electrodes are the most commonly used wet electrodes for bioimpedance measurements, also known as gel electrodes, as shown in Fig. 6(a). This type of electrode has the advantages of low contact impedance, stable measurement signal, high signal-to-noise ratio, and avoidance of cross-infection, making it the preferred and standard choice for electrophysiological measurements in clinical and research settings [58], [59], as well

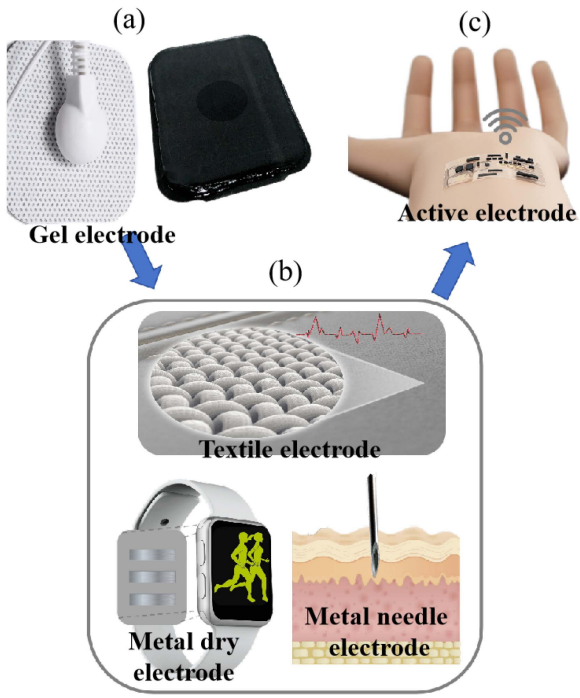

FIGURE 6. (a) Wet electrode; (b) Dry electrodes; (c) Active electrode.

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as the most commonly used electrode in bioimpedance measurements. However, Ag/AgCl electrodes have been shown to have several disadvantages in scenarios where they are used for wireless, mobile and long-term monitoring [58], [60], [61], [62]: 1) the dependence on electrolytes can lead to signal quality degradation due to dehydration of the gel during prolonged use; 2) the required spacing between electrodes may be too small, and the overflow of electrolytes may cause short-circuiting of bioelectrical signals; 3) there are toxicological problems associated with the electrolyte gels, which may cause skin irritation or allergic dermatitis; 4) the electrodes do not fit perfectly on the skin and the measurement signals show motion artifacts when the patient moves. Therefore, in the last decades, researchers have tried to develop reliable and effective alternatives to electrodes for recording biopotentials for physiological monitoring. In contrast, dry electrodes, such as rigid metal dry electrodes [63], soft rubber-based dry electrodes [64], sponges [60], and fabrics [65], [66], [67] are more versatile and reusable. Rutkove et al. [68] pointed out that the use of conventional electrodes for EIM in the clinic requires repeated application and removal of the electrodes, which is very inconvenient and time-consuming. Therefore, they developed a stationary handheld electrode array (HEA). The HEA has four stainless steel electrodes corresponding to the current and voltage electrodes and is reusable with saline-moistened skin. They performed repeatability tests with the HEA and showed that the HEA was reproducible with an intraclass correlation coefficient as high as 0.99 and that the HEA data correlated closely with the adhesive electrode data [68]. Li et al. [29] used the HEA array to measure the change in EIM of 19 healthy subjects during a fatiguing task and found that fatigue resulted in a decrease in resistance. Both examples illustrate the applicability of HEA for EIM measurements and greatly improve efficiency by eliminating the tedious process of attaching electrodes. Retractable metal electrodes are also available, and the advantage of this class of electrodes is that they are easy and quick to use [69]. However, the rigid metal electrodes do not fit closely to the skin and need to be pressed against the skin by an external force, which is not suitable for the mobile monitoring scenario of EIM. In addition, the gap between the rigid electrode and the skin leads to large motion artifacts with high skin contact impedance [70], [71]. In 2000, researchers selected three dry electrodes made of different materials (stainless steel, titanium, and aluminum) to compare with Ag/AgCl electrodes. It was found that the average contact impedance of the dry electrodes was twice as high as that of the wet electrodes [58]. At the beginning of the measurement, the dry electrodes were more affected by motion artifacts. However, the contact impedance of the dry electrode under the influence of motion artifacts decreases with time and reaches the same level as that of the wet electrode or even exceeds it. The dry electrode is less affected by moving charges, namely 100 times less than the wet electrode [58]. The reason for the better performance of the dry electrodes over time is that the sweat acts like an adhesive [72]. Therefore, dry electrodes will perform better than wet electrodes in scenarios such as pursuing time efficiency or being subject to high electromagnetic field interference.

For wearable devices, reliable long-term signal acquisition and user comfort are critical. Flexible electrodes are an effective solution for rigid materials that do not conform to the skin, such as dry textile electrodes and rubberized flexible electrodes, many teams in this field have addressed. Javier Ferreira et al. [69] have customized a portable four-electrode bioimpedance spectrometer equipped with textile electrode suits for the assessment of body fluids in dialysis patients. The textile electrodes were made from a biocompatible, conductive, silver-based Shieldex P130 + B fabric from Statex, with the conductive fabric coated with 99% conductive silver. The textile dry electrodes shown in this paper have very low contact impedance, which differs from other textile dry electrodes. The main reason for this is the large contact area of the electrode, which is ten times larger than that of a standard Ag/AgCl electrode [65], and the experimental results have proved that the textile dry electrode is reliable in measuring the impedance spectra. Wang et al. [66] compared the performance of five silver-coated textile ribbon electrodes with that of conventional Ag/AgCl electrodes. The results showed that the ribbon electrodes had a more uniform current distribution than the traditional electrodes but still suffered from motion artifacts. Several other teams have reported on various dry textile electrodes for bioimpedance applications [66], [73], [74], [75]. In summary, Ag/AgCl electrodes are still the most reliable electrodes for measuring bioelectrical signals, but they are not convenient for clinical use and are not suitable for wearable and mobile scenarios. Flexible electrodes are currently the most suitable electrodes for sports and wearable designs. Their comfort is the most acceptable for the user. However, several problems need to be solved: 1) the high contact impedance between flexible electrodes and the skin as well as motion artifacts may lead to serious measurement errors; 2) sweating may cause significant changes in contact impedance, resulting in inconsistencies in measurement data before and after measurement; 3) whether the flexible electrode can be washed is related to the reliability of reusing dry textile electrodes; 4) the most critical point is that the flexible electrode must also have the function of wireless data transmission or wireless communication.

All electrodes described above are passive electrodes. Even though we have classified the electrodes as dry or wet electrodes, a further distinction contributes to a clearer understanding of the electrical behaviour of the electrodes. Active electrodes are passive electrodes with active devices such as intrinsic buffers or amplifiers added [76]. Yi Wang et al. [77] developed a flexible active electrode for electrical impedance imaging in 2023. The electrode contains a three-layer structure: the first layer of the island bridge structure integrates the excitation measurement circuit; the second layer of the serpentine copper wire structure connects the integrated circuits; and the flexible electrode is attached to the surface of the object to be measured, and is contacted by a circular electrode. It is confirmed that its deformation does not affect

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the signal, and it has higher accuracy and anti-interference compared to conventional electrodes, making it suitable for real-time skin monitoring [77]. Fig. 6(c) shows a flexible active electrode. The issues of electrode contact impedance and motion artifacts were not tackled in that work. As the demand for mobile health monitoring is gradually increasing, the need for wearable biosensing electrodes that are lightweight, flexible, and can be continuously monitored is increasing, making integrated active electrodes an important direction for future development. For example, electronic textiles are the next development direction of the dry textile electrode mentioned above. This means that biocompatible and biostable electronic circuits are worn on the textile to realize the recording and transmission of physiological signals.

Most EIM studies have used noninvasive measurement electrodes, including the wet, rigid, dry, and flexible electrodes mentioned above. These types of electrodes are more acceptable and have the advantages of high conductivity, noninvasiveness, painlessness, and safety. However, electrodes in direct contact with the skin are significantly affected by electrode-skin contact impedance, motion artifacts, and noise, which undoubtedly sacrifice some of the detection accuracy compared to invasive detection. In 2016, Zhao Li et al. [78] proposed the use of microneedle electrode arrays (MEA) to overcome the problem of electrode-skin contact impedance, making them useful for the characterization of neurogenic myelopathies. Experimental results showed excellent reproducibility of MEA with intraclass correlation coefficients above 0.920. In 2018, the researchers used finite element modeling to verify the feasibility of novel electrical impedance imaging needles to visualize healthy and diseased skeletal muscles at the microscopic level [79]. In 2021, the team also proposed using concentric EIM needles to record muscle impedance, with the two electrodes at the tip of the needle being current- and voltage-sensing electrodes [80]. Needle electrodes eliminate the influence of skin and subcutaneous fat tissue to a certain extent compared to adhesive electrodes. This is particularly important for patients with excessive obesity, but the discomfort experienced by patients during use must be considered. In addition, there may be an opportunity to use microneedle arrays for long-term monitoring of muscle status based on needle electrodes. This method reduces the discomfort of conventional needle electrodes and eliminates the effects of contact impedance, resulting in a higher-quality bioelectric signal.

D. MODELING AND ANALYSIS METHODS

1) EQUIVALENT CIRCUIT MODELING

The use of equivalent circuit modeling of biological tissues provides a reliable method of understanding the relationship between EIM measurement data and muscle morphology and physiology [80]. The cell membrane is primarily composed of a phospholipid bilayer and proteins, with the lipid bilayer interior filled with dielectrics. The extracellular and intracellular fluids outside and inside of the cell membrane, respectively, can be treated as conducting solutions. The cell membrane composed of the lipid bilayer can be electrically modelled as a capacitor [81], [82]. When direct or low-frequency alternating current is applied to biological tissue, the current predominantly passes through the extracellular fluid, bypassing the cell. As the frequency of the applied signal increases, the capacitive impedance of the cell membrane decreases, allowing a portion of the current to flow through the membrane capacitance and intracellular fluid. As the frequency increases, the capacitive impedance of the cell membrane gradually decreases. In the ideal state, the biological tissue will reach the minimum impedance value when the frequency tends to infinity, as shown in Fig. 7(a). Consequently, biological tissues exhibit higher impedance at low frequencies and lower impedance at high frequencies, effectively illustrating the capacitive properties of cell membranes. Biological tissues, comprising numerous cellular circuit models connected in series, can be represented by a three-element circuit, as shown in Fig. 7(b). Lukaski et al. [83] experimentally supported this theory by assessing the R and reactance (X) at a frequency of 50 kHz of uniformly sized and shaped potatoes before and after cooking. The results demonstrated a 100% decrease in X and a 54% decrease in R after cooking [83], indicating that heating disrupts the cell membrane, leading to the loss of membrane capacitance, while the decrease in R is attributed to the expansion of extracellular fluid after the rupture of the cell membrane. Therefore, the equivalent formula for bioimpedance can be expressed as:

$$Z = \frac{R_e(1 + j\omega C_m R_i)}{1 + j\omega C_m(R_e + R_i)} \tag{1}$$

where Z is the impedance value of the biological tissue, R_e is the equivalent resistance of the extracellular fluid, R_i is the equivalent resistance of the intracellular fluid, $j = \sqrt{-1}$, and ω is the angular frequency.

It assumed that:

$$\begin{aligned} \tau &= (R_e + R_i)C_m \\ R_0 &= R_1 \\ R_\infty &= \frac{R_e R_i}{R_e + R_i} \end{aligned} \tag{2}$$

which results in the following equation:

$$Z = R_\infty + \frac{R_0 - R_\infty}{1 + j\omega\tau} \tag{3}$$

where R_0 is the equivalent resistance at zero frequency, R_∞ is the equivalent resistance at infinite frequency, and τ is the time constant or relaxation time. According to the electrical properties of biological tissues, K. Cole concluded that the electrical impedance of biological tissues can be represented by a segment of an arc in the complex plane [84]. However, the actual measured value of bioelectrical impedance does not always appear as a semicircle with the center of the circle on the x-axis. Instead, only a segment of the arc in the fourth quadrant may be present, as illustrated in Fig. 7(d). In this figure, f_c denotes the frequency at which the maximum bioelectrical impedance value was obtained. To account for the deviation from the idealized semicircular pattern, the Coles

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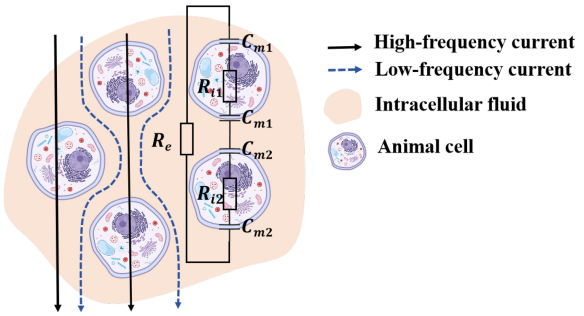
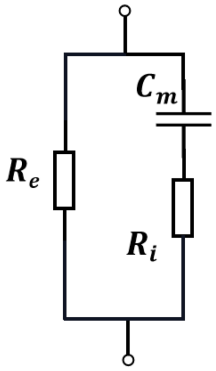
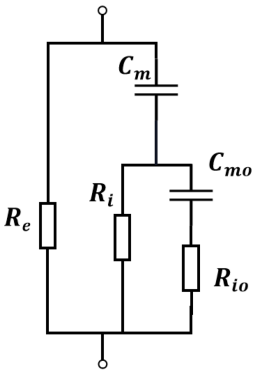


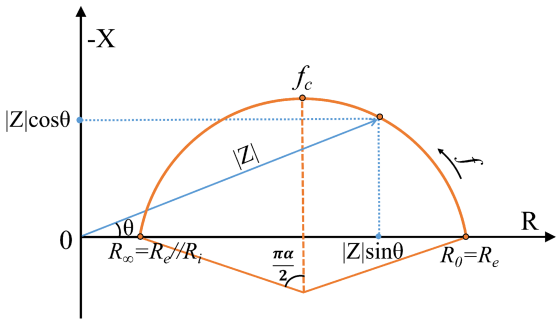
(a) Electric current path. At low frequencies, the current flows mainly through the extracellular fluid; at high frequencies, the current flows through the intracellular and extracellular fluids.



(b) Three-element equivalent circuit. R_e is the equivalent resistance of the extracellular fluid, R_i is the equivalent resistance of the intracellular fluid, and C_m is the membrane capacitance.



(c) Five-element equivalent circuit. C_{mo} is the organelle membrane, R_{io} is the intracellular fluid.



(d) Complex impedance trajectory diagram.

FIGURE 7. EIM circuit equivalent principle.

introduced modifications to the calculation model, resulting in the development of the Cole-Cole theory [85]. This theory expresses the bioelectrical impedance characteristic equation as follows:

$$Z(\omega) = R + jX = R_\infty + \frac{R_0 - R_\infty}{1 + (j\omega\tau)^\alpha} \tag{4}$$

where α is the scattering coefficient representing the position of the center of the arc, and it takes values between 0 and 1.

The exploration of extracting parameters from bioelectrical impedance data using bioimpedance equivalent circuits, and their subsequent application in analyzing the physiological state of biological tissues, has become a focus of research in [18], [86], [87]. The discussion of the model parameters to be measured is explicitly presented in the following subsection. The use of bioimpedance equivalent circuits to explain the changes in the measured tissue proves to be relatively straightforward. This method facilitates the interpretation of state changes, including pathological changes, in biological tissues by correlating these changes with variations in electrical parameters. Clemente et al. [88] used an equivalent circuit model to analyze the forearm flexor muscle in three states (rest, sustained contraction, and post-contraction phases) and verified that the circuit model was sensitive to changes in muscle state. Similarly, Kusche et al. [27] analyzed the electrical properties during muscle contraction by constructing an equivalent circuit for the forearm muscle. Despite the widespread adoption of the three-element circuit model, its limitation in adequately fitting multi-frequency EIM data over a wide frequency range has been acknowledged [89]. For this reason, in a series of three articles, Shiffman et al. discussed the applicability of the five-element circuit model for different populations, including healthy individuals, individuals with neuromuscular diseases, and individuals with fractures [89], [90], [91]. They replaced one of the resistors of a three-element with an additional three elements, thus constructing a more refined five-element circuit, as shown in Fig. 7(c). This modification took into account the presence of membranes and internal fluids in the cells. As expected, the five-element circuit provided more accurate predictions than its three-element counterpart [89], [90], [91]. In [92], an even more precise circuit model based on the anatomical structure of biological tissues was proposed. However, it has been observed that complex models, while accounting for detailed anatomical features, do not necessarily help to identify characteristic parameters accurately.

2) ELECTROMAGNETIC SIMULATION MODELING

The results of EIM measurements offer a phenomenological explanation for the changes in the electrical properties of the human body. However, they are not sufficient to provide a quantitative explanation of the effects of changes in non-uniform tissue morphology and electrical properties on current conduction within biological tissues. In addition, the existing studies lack scientific clarification of the role of EIM in characterizing muscle movements. Currently, finite element modeling analysis is proving to be a highly effective and feasible method for interpreting the observations made by EIM, especially in conjunction with variations in muscle geometry and electrical properties of tissues [93]. Most existing finite element modeling analyses of EIM are based on anatomical data used to reconstruct human tissues or organs. This includes data from various sources, such as ultrasound, magnetic resonance imaging (MRI), and computed tomography (CT). Electrical parameters of muscle tissue layers

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are usually obtained from animal measurements and publicly available databases, such as Gabriel's public database [94] and other online resources [95]. In 2012 Rutkove [95] constructed a finite element model of the human upper arm based on anatomical data, with electrical parameters of the tissue layers obtained from published databases and actual measurements of rat muscles. The study systematically quantified the effects of variations in muscle size, conductivity, and subcutaneous fat thickness on muscle impedance measurements using the finite element model [95]. Given the sensitivity of EIM to the electrode configuration of the measurement system, finite element analysis is also commonly used to evaluate the effects of electrode configuration on muscle impedance measurements and minimize this effect [95], [96], [97]. In addition, finite element modeling provides a non-invasive approach to investigate changes in the physical morphology of muscles in disease states [98], [99]. Wang et al. [98] developed a finite element model of the electrical parameters of rat muscles with different degrees of nerve damage. They aimed to quantify the state of deeper muscle tissues by measuring muscle impedance and limb shape, thereby providing new non-invasive insights about the state of diseased muscles in humans. In a word, finite element modeling has proven to be one of the most effective methods for investigating the biophysical mechanisms underlying EIM changes. It allows the researcher to noninvasively and quantitatively investigate the effects of various factors on muscle impedance measurements. In cases where physiological or microscopic changes do not provide adequate explanations, finite element modeling analysis can provide additional insights into the obtained EIM data.

It is worth noting that Alejandr et al. found in a comparative study that finite element modelling of the limbs was more accurate than three-element circuit modelling. However, the three-element circuit model demonstrated a significant advantage in terms of computational time [100]. In general, there has been a long discussion about circuit modelling, which is indeed difficult to match EM modelling in terms of accuracy. Moreover, with the rapid development of arithmetic resources, the computational complexity faced by EM modelling is no longer a major obstacle to its development.

E. PARAMETER REPRESENTATION AND DATA PROCESSING METHODS

1) *R*, *X*, f_c, ANISOTROPY

Conventional EIM parameters consist of impedance (Z), R, X, and phase (θ). Among them, Z denotes the impedance of the tissue under test to the current, R represents the impedance of the intracellular and extracellular ionic solutions to the current, X indicates the additional current impediment generated by reactive elements such as cell membranes, connective tissues, etc. And θ characterizes the geometric angle between R and X [101]. The relationship between the parameters is given by:

$$Z = R + jX = |Z|\, arg(\theta) \tag{5}$$

$$|Z| = \sqrt{R^2 + X^2} \tag{6}$$

$$\theta = \arctan\left(\frac{X}{R}\right) \tag{7}$$

Since the whole tissue under test is equivalent to a three-element circuit, the resistance at a frequency close to 0 (R_0) represents the equivalent resistance of the extracellular fluid, and the resistance at a frequency close to infinity (R_∞) can be considered as the intracellular resistance in parallel to the extracellular resistance. This conceptualization allows the extrapolation of the equivalent circuit parameters of biological tissue and provides insights into localized swelling, dehydration, muscle damage, and body composition analysis [84], [85]. The frequency corresponding to the maximum reactance is called the center frequency f_c, and in some studies, the phase angle is used as an alternative measure ($\omega_c = 2\pi f_c$) [102]. This parameter provides essential information about the relative density of the tissue [103], which has a linear relationship with the relaxation time. The parameter α is used to characterize tissue homogeneity and takes a value between 0 and 1. A value of $\alpha = 0$ indicates a perfectly homogeneous tissue, while a value of $\alpha = 1$ indicates a perfectly heterogeneous tissue [104], implying that there is significant variability within the tissue in terms of its mechanical or electrical properties, or other characteristics. Consequently, multifrequency EIM provides more comprehensive information, incorporating equivalent resistance of intracellular fluid (R_i), resistance value at frequency 0 (R_0), equivalent membrane capacitance value (C_m), f_c, and the four parameters in the characteristic equation (τ, R_0, R_∞, and α) in addition to the conventional parameters. In a 2009 study, Tao Dai et al. [104] reported that the f_c is the most reliable parameter for characterizing the electrical properties of biological blood. Furthermore, Qiao et al. [105] applied the bioimpedance technique to breast cancer suspension cells and found significant differences in bioimpedance parameters (f_c, α), which may indicate the cancerous nature of the cells.

Muscle anisotropy, which refers to the orientation-dependence nature of muscle impedance, is a remarkable property in which muscles oriented parallel to the muscle fibers exhibit higher conductivity than those oriented perpendicular to the muscle fibers. This phenomenon indicates that the current flows in the direction of the muscle fibers rather than across them [106]. This property is more commonly used in disease development studies. Rutkove et al. [107] found that anisotropy can be used to distinguish whether a muscle diseases is myopathic or neurogenic. In patients with amyotrophic lateral sclerosis (ALS), muscle anisotropy was increased and distorted, in contrast to the expected or decreased anisotropy observed in patients with myopathies [107]. Li et al. [17] applied the EIM technique to evaluate the changes in paralyzed muscles after a stroke and found that the electrical resistance and impedance of paralyzed muscles were significantly lower than those of healthy contralateral muscles. However, it is important to note that the anisotropy ratio

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is affected by factors such as electrode size, electrode spacing, muscle size, muscle shape, and electrode orientation relative to the muscle fiber [108], [109]. Therefore, the specific application conditions must be carefully considered and qualified when using this parameter. This also indicates that the reproducibility of muscle anisotropy is not consistently reliable.

2) DATA-PROCESSING METHODS

With the improvement of computer processing power and the rapid evolution of machine learning, data-processing methods are becoming more diversified. Machine learning techniques have demonstrated the advantages of efficient, automated analyses and deeper mining of hidden patterns and regularities in data processing, with particular excellence in handling nonlinear relationships. Therefore, using algorithms for feature discovery and deep data analysis will also become one of the main research trends in EIM data analysis. This section aims to overview the data processing methods and the application of machine learning techniques in EIM analyses, to provide useful insights and guidance for this research direction.

Data dimensionality reduction techniques, such as principal component analysis and tensor decomposition, are increasingly used in EIM studies to reduce data complexity while maximising the retention of useful information. Zagar et al. [110] performed a principal component analysis of the impedance spectrum in the range of 300 Hz to 75 kHz in 32 samples. They used clustering to distinguish between muscle contraction and relaxation and achieved a classification accuracy of 80%. In recent years, tensor decomposition has become widely used in data mining. It offers advantages over methods such as principal component analysis (PCA) and the L2 paradigm, especially when dealing with missing data. In 2021, Chlöe N. Schooling et al. [111] used non-negative tensor factorization (NTF) to characterize high-dimensional EIM spectra. They extracted three spectral modes and quantitatively analyzed their contribution in patients with amyotrophic lateral sclerosis as well as in a healthy cohort, and termed this approach tensor EIM. The results indicated that the data processed with tensor EIM showed higher sensitivity in distinguishing the patient group from the healthy group compared to the L2 paradigm processed features [111]. One year later, they continued to monitor the progression of extended medullary disease in amyotrophic lateral sclerosis using tensor EIM. They reduced the 168 data points per subject to a single value and found that tensor EIM was able to track and quantify disease progression with greater sensitivity to disease change than single-frequency EIM [112]. While tensor decomposition has been used in several biomedical signal studies, its combination with EIM data is very promising for disease analysis. This pioneering work represents a positive step forward by introducing valuable considerations for data analysis and application of EIM.

Machine learning techniques have long been used to analyze medical data and are now an indispensable tool in medically assisted diagnosis [113]. In the field of EIM, the integration of machine learning for muscle status detection and prediction is still in its infancy, focusing primarily on data mining, classification, and prediction of unknown data. Srivastava et al. [114] first introduced machine learning algorithms to neuromuscular diseases classification in 2012. By combining features from quantitative muscle ultrasound (QMU) and EIM techniques, they accurately classified muscles affected by spinal muscular atrophy using support vector machines. The results showed a classification accuracy that exceeded that of either EIM or QMU alone. Machine learning algorithms can also use EIM for non-invasive prediction of muscle fiber size and connective tissue deposition, contributing significantly to the "virtual biopsy" technique of EIM [115], [116], [117]. Our research team has recently implemented a mapping from EIM to grip force using long short-term memory (LSTM) networks that achieved an R-squared value greater than 0.9023. This promises to provide a new solution for the control of prosthetic limbs [118]. In addition, teams are also using the data mining capabilities of machine learning to predict total muscle mass by extracting electrical resistance, phase, and other features from the EIM [119]. This approach represents an attempt to self-test muscle mass in a non-medical setting. The development of machine learning has revitalized various fields, for example, autonomous diagnosis based on machine learning not only improves diagnostic accuracy, but also has the potential to reduce the burden on healthcare systems. Although the research of EIM combined with machine learning algorithms is in the experimentation stage, we can expect that machine learning algorithms will provide more possibilities for applying EIM. Especially in sports exercise, the combination of machine learning will improve the accuracy and convenience of muscle state self-assessment.

III. STATE OF RESEARCH

A. DETECTION OF MUSCLE CONTRACTION

The detection of muscle contraction is not only important in the biomedical field, but is also widely used in the field of human-computer interaction. In a study conducted in 2013 by Shiffman et al. [8], the impedance values of the anterior part of the forearm were measured at the frequency of 50 kHz. They found that the resistance and impedance of the finger flexors increased during isometric contraction and were nonlinearly related to the force value. They also found that the impedance values changed a few milliseconds before the force was applied and did not return to their original values after muscle relaxation. This demonstrates the high sensitivity of the EIM technique to changes in muscle status and proves that changes in muscle morphology alone do not exclusively determine EIM changes. Freeborn et al. [26] asked subjects to complete centrifugal exercise on one arm. In contrast, the other arm was used as a control without any exercise to compare the changes in EIM of the biceps brachii muscle between the two arms over 96 hours after exercise. They found that the electrical resistance and reactance values of biological tissues decreased after centrifugal exercise, with no statistical difference in time for the unexercised muscle, validating that bioimpedance techniques are sensitive to post-exercise changes and maybe a

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potential method to non-invasively quantify changes in tissues due to exercise, fatigue, injury and recovery. Kusche et al. [27] identified specific phase changes during muscle contraction that could serve as markers for detecting muscle activity. Furthermore, they made a notable attempt in [28] to combine EIM with EMG signals to achieve a more reliable detection of muscle contraction. In robustness tests involving mechanical disturbances, EIM was found to be more reliable than electromyography (EMG) for muscle contraction detection [28]. The collective results of the above studies show that EIM is a sensitive and reliable method for muscle detection. It is a promising biomarker of muscle activity with potential application in rehabilitation clinics, human-computer interaction, and exercise monitoring.

B. MONITORING OF MUSCLE FATIGUE

Muscle fatigue is a physiological phenomenon characterized by a decrease in muscle strength and a slowing of contraction to the point of muscle being unable to maintain a particular movement [120]. Fatigue induced by prolonged exercise can lead to impaired muscle function, e.g. hypoxic conditions and the formation of free oxygen radicals leading to increased lysosomal activity, as well as mechanical stress on fibres caused by high-intensity eccentric exercise, and so on [121]. In [29], Li et al. investigated changes in EIM induced by different levels of exercise and fatigue. They evaluated impedance changes in 19 healthy subjects during different levels of isometric muscle contraction (20% MVC, 60% MVC, and 100% MVC) as well as a fatigue task, where the MVC (maximal voluntary contraction) corresponded to the subject's maximal elbow flexion value measured with a torque device. Measurements were performed using HEA at 50 kHz and 100 kHz frequencies on the biceps brachii muscle. The results showed a significant increase in R at 60% MVC and 100% MVC compared to initial values (10.1% and 9.2%, respectively), which is consistent with Shiffman's experimental results [8]. The R decreased significantly when the muscles were fatigued (post hoc test, $p = 0.011$). They suggested that it was due to the accumulation of metabolites and intracellular fluids during muscle contraction, which increased the electrical conductivity of the muscle. However, the X, did not change significantly throughout the experiment. They concluded that muscle contraction did not induce structural changes in the muscle fibers, and therefore, the membrane capacitance did not change much. In 2018, Freeborn et al. [18] repeated the study of exercise-induced fatigue. Their results were consistent with those of Li et al. [28] and showed a decrease in R after fatigue. However, the X, which is determined by membrane capacitance, showed variations before and after fatigue. This difference was probably due to a fatigue protocol that differed from that used in [29], where participants performed repeated elbow flexion while holding a dumbbell. Impedance values were measured using an Agilent E4990 A impedance analyzer. In their study on changes in bioimpedance in mice before and after injury, Sanchez et al. [122] indicated that capacitance is more sensitive to muscle damage. Therefore, it is reasonable to assume that the difference in X between the two studies is due to the different levels of fatigue induced by the respective fatigue protocols, resulting in different degrees of muscle damage. In 2019, Freeborn [30] continued the study of impedance changes in the biceps muscle before and after fatigue, using an experimental protocol similar to that of [18]. The experimental results were also in agreement with those of [18], showing a significant decrease in R, X, and θ with increasing fatigue at a frequency of 50 kHz. While Freeborn et al. did not measure the changes in muscle swelling during their experiments, Yasuda et al. [123] used ultrasound imaging to confirm that muscle swelling increased with each additional set of biceps curl exercises. Therefore, Freeborn concluded that muscle swelling led to an increase in local tissue fluid, which in turn decreased the electrical impedance of the measured tissue. Huang et al. [31] measured the changes in electrical impedance parameters during dynamic contraction of the biceps brachii muscle in eight healthy subjects at different load levels (20% MVC, 40% MVC, and 60% MVC). They decided to validate the accuracy of the EIM application for estimating muscle fatigue using electromyographic signals. They used an EIM measurement system, different from any of the above studies, that included a signal generator, a constant current source, an Agilent 1141 A differential probe, and an Agilent MSO7054 A oscilloscope. Participants performed repetitive lower arm flexions and extensions while holding a dumbbell to the point of exhaustion. The main focus of the study was on the changes in R, without addressing the changes in X. The results showed a linear decrease in R at different load levels with increasing duration of contraction exercise. Also, there was a significant difference between the R values before and after fatigue in all 8 subjects ($p < 0$). The median frequency (MF) of surface EMG signals decreased faster with higher loads, indicating a faster onset of muscle fatigue. Notably, the decreasing patterns observed in both R and MF were consistent, providing further confirmation of the feasibility and reliability of EIM in assessing muscle fatigue. Zhou et al. [124] used EIM to assess muscle fatigue induced by neuromuscular electrical stimulation (NMES) and performed a comparative analysis using EMG features. Their results demonstrated that, following muscle fatigue, both the amplitude and phase of the impedance decreased compared to the relaxation state, but the phase angle θ showed instability. Both parameters exhibited a consistent decreasing trend as the mean power frequency of the surface EMG throughout the fatigue process. These results from Zhou et al. confirmed that the physiological mechanisms underlying muscle fatigue, whether induced by voluntary exercise or external stimuli, maybe the same. This observation could explain why various studies, despite utilizing different exercises and fatigue protocols, observed the same EIM changes, suggesting the reliable monitoring ability of EIM in the assessment of muscle fatigue.

The results suggest that a rapid decrease in impedance values during muscle contraction may serve as an early warning signal of potential muscle damage. Therefore, continuous monitoring during exercise is crucial to prevent muscle damage and maintain optimal physical condition. The discussion in this subsection confirms that EIM is a biosensor capable

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of accurately identifying muscle status and has the potential to be used as a wearable tool for continuous monitoring of muscle status during exercise. Furthermore, EIM proves to be a promising solution for bionic applications and intelligent rehabilitation.

C. ASSESSMENT OF MUSCLE INJURIES

Resistance of EIM signals primarily depends on the intracellular and extracellular fluids, while reactance value is related to the condition of the cell membrane. It is expected that EIM has the potential to detect damage to the cellular architecture of biological tissues and variations in tissue fluids resulting from injury. In the context of EIM research, most studies have focused on its application in the assessment of neuromuscular diseases and its ability to monitor disease progression. However, there is a limited number of studies addressing the use of EIM in the context of muscle injury. As part of this review, we have presented several studies that highlight the importance of EIM in assessing, classifying, and tracking wound healing in the context of muscle injury. In 2012, Lukaski and Moore [32] used EIM (referred to as localized bioelectrical impedance analysis, L-BIA) to monitor the wound healing process. Using a phase-sensitive instrument (Quantum IV, RJL Systems; Clinton Township, MI) they recorded R, X, and θ at 50 kHz. Four small adhesive electrodes (TYCO Healthcare. Mansfield, MA) were placed on the skin and near the edges of the wound. The results showed that R, X, and θ gradually increased as healing progressed, but transiently decreased at the onset of erythema during wound healing. In another case of implant repair of a surgical incision for ligament repair, wound debridement caused a transient decrease in R, X, and θ (2%, 28% and 30%, respectively), suggesting significant disruption of cell membranes and function with minimal changes in extracellular fluid. Tissue trauma associated with wound preparation and implant placement resulted in decreases in R, X, and θ (23%, 65% and 59%, respectively), followed by significant increases in R, X, and θ (28%, 210% and 178%, respectively) after recovery from the implantation procedure. These observations indicated that during wound healing, changes in the cytosol were limited, whereas significant differences in X highlighted variations in cell numbers and wound repair capacity. A decrease in X indicated damage to the cellular structure, while an increase indicated increased cell numbers as a result of wound healing. In addition, in [32] the authors also found that a decrease in R, X, and θ predicted the onset of infection even before the post-infected wounds were examined by laboratory methods. This potential early warning sign of wound infection could allow more time for intervention in severely injured patients. It is important to note that the reference had reported a limited sample size, with examples consisting of only one case and lacking information about the injured persons. However, similar results were reported in [32] and [33]. In [33] EIM parameters and histologic status were measured on a fresh piece of beef before and after heating in a microwave oven for four minutes. The heated beef showed a decrease of 74.9%, 95.6%, and 80.9% in R, X, and θ, respectively. Histological analysis showed damage to connective tissue, and disruption of cell membranes and extracellular matrix (gap-filling), which is consistent with the suggestion made in [32]. Therefore, EIM shows promising potential in predicting wound healing and tracking changes in wounds, justifying further research.

The lower limbs are a common site of muscle injury in football players. Lukaski et al. [34] investigated the changes in L-BIA (i.e. EIM) parameters of the quadriceps, hamstring and calf muscles of three male football players before, during and after injury and recovery. Using MRI images as a reference, they investigated the relationship between the different levels of muscle injury and L-BIA parameters. The results showed that L-BIA parameters decreased depending on the severity of muscle injury. Grade I injuries showed changes in the values of R by 11.9%, X by 23.5%, θ by 212.2%; grade II injuries showed changes in the values of R by 20.6%, X by 31.6%, θ by 13.3%; and grade III injuries showed changes in the values of R by 23.1%, X by 45%, and θ by 27.5%. The results showed that a decrease in X mainly reflected changes in muscle structure after injury, whereas a decrease in R predicted tissue edema (localized fluid accumulation). The changes in L-BIA during muscle recovery to RTP were consistent with the wound-healing process reported in [29]. However, similar to the previous reference [29], the study in [34] also had a limited sample size with only three cases. Extending the research protocol of [34], another study [35] continued the L-BIA analysis of the severity of the three lower limb muscle injuries in professional football players. It included a total of 21 injury cases (grade I: 11, grade II: 8, grade III: 2), with measurements taken 24 hours after injury. Using the L-BIA values measured on the uninjured leg on the symmetrical part of the injury side as baseline values, it was found that R, X, and θ showed a decreasing trend after the injury. The average decreases for the three injury levels were grade I: 10.4%, 17.5%, and 9.0%; grade II: 18.4%, 32.9%, and 16.6%; and grade III: 14.1%, 52.9%, and 43.1%, respectively. Interestingly, the change in R was not proportional to the severity of the injury. This change in R was confirmed by MRI, where in one case the contents at the grade III injury were not fluid (blood or edema) but a semi-solid cystic hematoma. Statistical analysis, including a one-way analysis of variance (ANOVA) test analysis of the results on the injured versus uninjured leg and a statistical analysis of variance using the Scheffé post-hoc test for the percentage difference in R, X, and θ between the three injury levels, showed that only X had a high degree of variability. The results, which were further strengthened by the larger sample size, suggested that X could be used as a stability marker for the degree of muscle damage in the lower limbs. In addition, a combined analysis of the decreased ratios of R and X could provide insights into the morphology of the injury contents.

The time required for an athlete to RTP after an injury is a significant concern in treatment. In [125] it was shown that the RTP depends primarily on the severity of the injury, which in turn is related to the involved part of the muscle. Therefore, in [36], L-BIA was used to predict RTP in professional football players by categorizing muscle injuries based on the

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severity and location of injury as determined by MRI. A total of 22 muscle injuries were categorized into three grades: grade 1 (n = 7), grade 2f (n = 8), and grade 2 g (n = 7), with the value at the contralateral uninjured site as the baseline values. The post hoc analyses showed significant statistical differences between X and θ in the pairwise comparisons across the three grades. With regard to RTP, a one-way ANOVA test ($F = 41.286$, $p < 0.000$) showed significant statistical differences for grades 1 (7 ± 2 days), 2f (14 ± 6 days), and 2 g (48 ± 15 days). The degree of muscle damage was proportional to the size of the gap, suggesting that the severity of the gap and muscle disruption negatively affected the time to RTP. Therefore, L-BIA may prove valuable in diagnosing and predicting time to RTP. In another study [36], L-BIA was used to analyze tendon, myotendinous junction (MTJ) and myofascial junction (MFJ) injuries [35], [125]. The L-BIA parameters showed decreases at all injury sites. Statistical analysis revealed that the L-BIA parameters were not sensitive to the anatomical changes due to the tendon injury, but could discriminate between MTJ and MFJ injuries. Of the three injury grades, only X showed significant differences for MTJ. No differences were found between MTJ and MFJ injuries in terms of RTP.

Consistent results were found by Li et al. [126] in the identification of traumatic peripheral nerve injury (TPNI) in the upper limb. In 39 patients with TPNI, EIM was measured in three muscles of the healthy and injured limbs, including the adductor pollicis brevis (APB), adductor digit minimi (ADM) and extensor digitorum (EDC), corresponding to median nerve injury (MNI), ulnar nerve injury (UNI), and radial nerve injury (RNI), respectively. The results showed a decrease in R, X, and θ on the injured side compared to the healthy side. Significant asymmetry of X and θ was observed in the muscles directly affected by the nerve injury, but not in R. The absence of considerable asymmetry of EIM parameters found in the unaffected muscles demonstrated the ability of EIM to differentiate TPNI.

In summary, EIM has excellent potential for the assessment of muscle injuries and the monitoring of wound healing. After the onset of injury, EIM parameters show a decrease that corresponds to the severity of the injury and gradually returns to the uninjured state during the healing process. However, in the studies described in this paper, R was found to be unreliable due to its susceptibility to measurement factors such as electrodes, whereas X was found to be highly sensitive and reliable in assessing injury. It is important to note that, due to individual differences, these studies could only use the healthy side of the injured person as a reference, limiting their application to bilateral injuries. Future research protocols should take this limitation into account. In terms of research development, studies with larger samples and longer duration will provide more reliable results. Furthermore, while studies have primarily focused on the 50 kHz EIM parameters, it is worth exploring multi-frequency EIM, which contains more physiologic information and may offer more comprehensive knowledge and applications. Despite these considerations, EIM remains a safe, cost-effective and practical method for assessing injury in professional athletes.

IV. FUTURE DIRECTIONS

Although EIM has been shown to have potential in clinical applications and muscle monitoring, there are still many challenges for its clinical use. Therefore, we propose the following research directions, which can be further developed based on the previous discussion.

A. WORKING MECHANISM OF EIM

The working mechanism of EIM in the assessment of muscle condition represents a crucial aspect of its theoretical development. Although numerous studies have aimed to establish correlations between the EIM and functional, electrophysiologic, imaging, and histologic outcomes and to compare them to current clinical assessment tools, a complete understanding remains elusive. These studies strongly confirm the clinical utility and value of EIM from a statistical point of view, however, there is still a lack of mechanistic exploration of the physical and physiological aspects of the EIM technique. Furthermore, a complete electrophysiological explanation for the relationship between EIM changes and muscle and underlying pathology is lacking [127]. While previous research has demonstrated that EIM is capable of predicting muscle fiber size and estimating muscle cross-sectional area [115], [116] circuit modeling and finite element modeling have predominantly taken a phenomenological approach. These approaches are designed to back-propagate subtle changes in the structure of biological tissues and fail to directly touch upon the intrinsic mechanisms by which EIM monitors changes in muscle state. The use of animal models may provide more convincing mechanistic explanations. For example, Sanchez et al. [128] observed significant differences in the dielectric spectra of slow skeletal fibers and fast muscle fibers between 1 MHz and 10 MHz, depending primarily on their mitochondrial content. However, whether it is for the immediate monitoring of short-term exercise effects or the tracking and evaluation of long-term muscle injuries, the current challenge is still the unclear correspondence between EIM parameters and specific physiological phenomena, muscle physiological functions and morphological changes. Therefore, future research needs to focus on building more refined models that concern the latest advances in molecular biology, biophysics, and clinical practice, to reveal the intrinsic mechanisms of EIM technology.

B. IMPROVEMENTS IN MEASUREMENT TECHNOLOGY

Improving measurement technology is essential for the practical application of EIM in various areas. Robust measurement methods are crucial, as the electrical impedance values of biological tissues are susceptible to various interferences caused by different experimental conditions and equipment, which can lead to inaccurate measurements. Stray capacitance, a common cause of high-frequency artifacts in bio-impedance measurements, has been identified as a major challenge. The "hook phenomenon", characterized by an increase in with

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reactance increasing frequency, particularly above 500 kHz, is observed in high-frequency measurements and is often attributed to stray capacitance originating from cable capacitance between the signal ground and the earth and the capacitance between the human body and the earth [129], [130], [131]. In addition, errors may arise from impedance mismatch and signal crosstalk during measurements [132]. This emphasizes the importance of identifying and eliminating sources of errors in EIM measurements, as reliable data sources are a prerequisite for conducting all studies. Reports from different research teams have shown that a wide range of biological tissue measurement devices have been used, leading to inconsistent experimental results. Therefore, it is crucial to calibrate the measurement devices or the measurements themselves to ensure consistency between devices. In addition, innovations in electrode technology are critical to improving technical accuracy and obtaining high-quality signals. As discussed in Section. II-C, each electrode type has its advantages and disadvantages for specific applications. For sports scenarios, the key requirements for electrodes are to minimise motion artefacts, ensure secure adhesion and enhance comfort. Currently, flexible active electrodes are expected to be the preferred solution to meet these requirements.

Indeed, existing measurement devices only meet the requirements of real-time sports monitoring to a limited extent. Therefore, solutions for miniaturization of the devices and mobile signal transmission are required. The well-being and comfort of the user/patient and the power consumption of the devices must also be taken into account. Overcoming these issues will significantly improve the comfort of personal applications and pave the way for a broader and more effective use of EIM technology.

C. STANDARDIZED MEASUREMENT PROCEDURES

Standardized measurement procedures have contributed to the EIM becoming a standard clinical tool. The EIM can be affected by the electrode material and is very sensitive to the electrode configuration. Therefore, many researchers have used finite element modeling to optimize electrode configurations to reduce the influence of other tissue layers and maximize the contribution of the muscle layer. However, different teams have built models and proposed new electrode configurations to suit their needs. This has not been conducive to the development of EIM, and the lack of uniform standards has led to researchers repeating the same procedures, resulting in inconsistent quality of research. Ahad et al. [96] compared the effects of electrodes with the same contact area but different shapes on EIM measurements and found that experimental results varied when different electrode shapes were used, with round electrodes improving the validity of EIM measurements. In addition to the location and shape of the electrodes, the posture of the human subjects also varied among reports. Some subjects were measured in a relaxed supine position [34], [133], [134]. Others, however, used a standing position that resulted in varying degrees of muscle contraction [135], [136], a major factor contributing to the lack of comparability of the existing studies. Of course, there were many factors that we did not explicitly list here, but it was undeniable that industry standardization is essential, including but not limited to the following points: 1) the development of standards for the use of electrodes, including the range of contact impedance, placement position, etc.; 2) the development of standards for electromagnetic safety and electrical engineering for equipment in response to the problem of the diversity of measuring equipment; 3) the standardization of measurement procedures to avoid differences in results due to different body positions, etc.; 4) the standardization of measurement procedures to avoid differences in results due to different body positions, etc. Standardized practice avoids duplication of work in a study and facilitates the exchange of results between researchers. EIM needs to become a standard clinical assessment tool, which requires joint efforts and constructive contributions from the entire research community in a sustained and comprehensive endeavour.

D. FUSION OF PHYSIOLOGICAL PARAMETERS

Changes in the human condition are very complicated, and assessing muscle status or clinical diagnosis from multiple perspectives will inevitably provide more comprehensive information. As noted by Roy et al., different techniques were sensitive to different pathologic features of the muscle [137], and other methods may capture different aspects of the biological tissue. The combination of EIM and ultrasound to describe muscle structure or to diagnose muscle diseases was the most common application [138], [139], [140], and the results of the combination were superior to those of either technique alone [114]. Additionally, integrating EIM with electromyography improved the accuracy of muscle contraction detection [28]. Briko et al. [141] proposed a bionic control method based on electrical impedance (EI), EMG, and force myography (FMG) signals, which was expected to enable anthropomorphic control of prostheses, orthotics, and rehabilitation devices. EMG reflects the electrical activity of the muscle itself, EIM reflects the contractile properties of the muscle, and FMG monitors the changes in skin morphology caused by muscle contraction [141]. EIM and other quantitative techniques can complement each other to assess muscle status from multiple perspectives comprehensively. In addition to the diverse fusion of detection techniques, the combination of EIM and machine learning will also drive progress in this field. With the improvement of computing power, machine learning has a significant advantage in data selection and data fusion, which also drives the rapid development of multimodal fusion. With the aid of machine learning, the applications of EIM are ready for rapid expansion.

V. CONCLUSION

In conclusion, Electrical Impedance Myography (EIM) emerges as a highly sensitive technology, adept at capturing muscle changes, including those induced by exercise. Its capacity to discern significant impedance parameter variations in response to muscle fatigue or sports injuries positions it as a valuable asset for injury identification and monitoring the subsequent healing process. Offering accurate feedback on

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muscle status, EIM also presents the potential to predict the Return to Play (RTP) of athletes.

TABLE 1. List of Abbreviations

Abb.	Definition
EIM	electrical impedance myography
RTP	return to play
R	resistance
OFDM	orthogonal frequency division multiplexing
HEA	handheld electrode array
MEA	microneedle electrode arrays
X	reactance
MRI	magnetic resonance imaging
CT	computed tomography
Z	impedance
θ	phase
ALS	amyotrophic lateral sclerosis
PCA	principal component analysis
QMU	quantitative muscle ultrasound
LSTM	long short-term memory
EMG	electromyography
MVC	maximal voluntary contraction
MF	median frequency
NMES	neuromuscular electrical stimulation
L-BIA	localized bioelectrical impedance analysis
ANOVA	analysis of variance
MTJ	myotendinous junction
MFJ	myofascial junction
TPNI	traumatic peripheral nerve injury
APB	adductor pollicis brevis
ADM	adductor digiti minimi
EDC	extensor digitorum
MNI	median nerve injury
UNI	ulnar nerve injury
RNI	radial nerve injury
EI	electrical impedance
FMG	force myography
DFT	discrete Fourier transform

The promising applications of EIM in the realms of sports and health suggest a paradigm shift, providing a viable alternative to expensive and cumbersome testing methods in professional athletes' activities. Simultaneously, EIM holds the promise of becoming an efficient tool for continuous monitoring in daily sports and fitness routines.

However, despite its potential, the current development of EIM in the field of sports and health calls for attention in several key areas. Firstly, a more scientific explanation of the mechanisms behind changes in EIM muscle feedback is crucial for comprehensive understanding. Secondly, advancements in detection accuracy and applicability to exercise fitness monitoring are essential to enhance its effectiveness. Thirdly, the standardization of measurement equipment and procedures is imperative for ensuring consistent and reliable results. Finally, the integration of other detection techniques can enrich the assessment process by providing a more comprehensive evaluation from multiple angles.

Addressing these considerations will undoubtedly contribute to the evolution of EIM into a more scientifically robust and widely accepted clinical assessment tool, thereby unlocking its full potential in enhancing sports performance and health monitoring.

Table 1 is the list of abbreviations.

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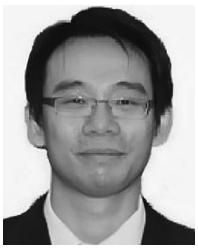

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