Portable Device for Angular Measurement and Muscle Activation of the Knee in the Sagittal Plane

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Abstract— **Gait alterations associated with conditions such as cerebral palsy, stroke, and knee osteoarthritis require accurate, accessible, and integrated assessment tools. This paper presents the design and validation of a portable device for the simultaneous measurement of knee joint kinematics and muscle activity in the sagittal plane. The system combines a Hall effect sensor for angular measurement and electromyography (EMG) sensors for muscle activity acquisition within a microcontroller-based architecture. A gait experiment over a complete cycle was conducted to evaluate system performance, focusing on the biceps femoris and rectus femoris muscles. The obtained EMG signals exhibited activation patterns consistent with physiological expectations: increased biceps femoris activity during the swing phase and peak rectus femoris activation during the stance phase. Additionally, the device measured knee joint angles within a range of 0° to 65°, achieving a Mean Absolute Error (MAE) of 2 $^\circ$ compared to conventional goniometry and a Mean Absolute Deviation (MAD) of 3.09°. These results validate the accuracy and reliability of the proposed system for capturing synchronized neuromuscular and kinematic data. The device offers a portable, low-cost, and user-friendly solution with potential applications in clinical gait assessment, rehabilitation monitoring, and biomechanical analysis.**

I. INTRODUCTION

Gait alterations associated with conditions such as cerebral palsy, stroke (CVA), and knee osteoarthritis represent a major clinical concern due to their high prevalence and their impact on mobility, joint kinematics, and overall quality of life. Cerebral palsy, a neurological disorder affecting movement, posture, and balance, presents a prevalence of 2 to 3.5 cases per 1,000 live births in developed countries [1]. Similarly, stroke shows an incidence of 255 cases per 100,000 individuals per year in Latin America and the Caribbean [2]. Knee osteoarthritis, a degenerative joint disease, affects 1.55% of the population in Peru and up to 37.4% of adults over 60 years of age in the United States [3]. These conditions frequently lead to abnormal gait patterns, including altered joint angles, reduced mobility, and impaired neuromuscular coordination, which require precise assessment for effective diagnosis and rehabilitation.

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Accurate characterization of gait involves not only the measurement of joint kinematics but also the analysis of muscle activation patterns. In this context, electromyography (EMG) provides valuable information on neuromuscular activity, while angular measurements enable evaluation of joint range of motion. The integration of both modalities is essential to achieve a comprehensive understanding of gait biomechanics and to support clinical decision-making processes.

Existing technologies for assessing knee biomechanics have evolved across commercial, academic, and patented solutions. Commercial tools such as universal goniometers, inclinometers, and mobile applications are widely used due to their accessibility and ease of use; however, they often suffer from limited accuracy, operator dependency, and increased costs in more advanced implementations [4].

Academic developments have introduced alternative approaches, including modified inclinometers, plastic optical fiber sensors, and portable 3D-printed goniometers, thereby improving portability, stability, and connectivity. Nevertheless, these systems still face challenges related to calibration, clinical validation, and long-term reliability [5], [6].

On the other hand, patented solutions incorporate advanced technologies such as inertial sensors (IMUs) and haptic feedback systems, enabling real-time measurement and therapeutic assistance. Despite their capabilities, these systems are typically associated with higher complexity, cost, and limited accessibility [7], [10].

Despite these advances, most available systems focus on either joint angle measurement or muscle activity analysis independently. The lack of integration between these modalities in portable, low-cost devices limits the acquisition of synchronized, comprehensive biomechanical data, which is essential for accurate gait evaluation.

Therefore, there is a need for an accessible and integrated system capable of simultaneously measuring joint kinematics and muscle activity. This work presents a portable embedded device that combines Hall-effect angular sensing with analog DFRobot EMG acquisition on a unified platform. The Hall-effect sensor was selected due to its high sensitivity, low cost, compact size (5 × 5 mm), and minimal weight (5 g), providing an efficient alternative to IMU-based systems [8]. Similarly, the analog DFRobot EMG sensors were chosen for their comfort, compact dimensions (22 × 35 mm), low weight (36 g), and integrated amplification and filtering circuitry [9]. The proposed system enables synchronized real-time acquisition, offering a low-cost and portable solution for gait assessment, clinical evaluation, and rehabilitation monitoring.

II. METHODOLOGY

A. System Requirements

The system includes a set of functional requirements focused on measuring knee biomechanics. Its main function is the measurement of knee angular values and muscle activity associated with movement in the sagittal plane. The system must accept input signals such as power on, start measurement, and end measurement. In turn, it must generate output signals in the form of graphs representing the knee's angular position (in degrees) and the electrical activity (in mV or µV) of selected muscles as a function of time [6].

From a kinematic perspective, the device must support knee flexion and extension between 0° and 140°, given that a common functional range during gait is 0° to 90°. Additionally, the system is designed with a single degree of freedom, enabling active range of motion (AROM), in which the patient performs the movement voluntarily without direct intervention from the examiner, who remains present only for support [12].

Regarding anthropometric requirements, the device must be adaptable to a range of user body characteristics. This is achieved through the use of compression garment technologies, such as Velcro straps or elastic bands, which provide a customized, comfortable fit for each user [13].

To ensure portability, user comfort, and minimal biomechanical interference during gait, the device weight was limited to under 1 kg, as masses exceeding this value may significantly alter natural movement patterns and increase energy expenditure [14].

It must also provide a minimum battery life of one hour to ensure uninterrupted operation. Furthermore, the setup and usage time should be approximately 15 minutes, aligning with the average duration of medical appointments in Peru [11] [15].

Safety requirements establish that the device must be non-invasive, ensuring minimal contact with the user's body. It must also comply with electrical safety standards, including appropriate current and power specifications for cables, as well as the integration of protection circuits [16].

Environmental requirements specify that the device should operate within a temperature range between 15°C and 27°C to prevent failures in electronic components [16].

Finally, maintenance and availability requirements indicate that the design must allow for easy replacement of parts, ensuring that components are readily available in the local market to facilitate maintenance and long-term usability [17].

B. Electronic and Mechanical Design of the Device

This section presents the mechanical and electronic design of the proposed device to facilitate a clear understanding of its operation. The system is divided into two main components: the lever arm and the casing.

1) Assembled Device: Fig. 1(a) illustrates the fully assembled device, including all required components, ready for the attachment of Velcro straps and placement on the participant.

2) Lever Arm: Fig. 1(b) shows the lever arm, a movable component that plays a key role in both securing the device to the participant and enabling the measurement of knee joint range of motion. A magnet is embedded in its circular section, allowing angular displacement to be captured. Additionally, the structure incorporates holes for Velcro straps, ensuring proper fixation to the participant's leg.

3) Front View: The front view of the device is presented in Fig. 1(c). This section includes the main user interface elements: a power switch, a digital voltmeter, and status LEDs. The switch controls device operation, the voltmeter indicates the battery level, and the LEDs provide visual feedback on the device's current state.

4) Internal Electronics: Fig. 1(d) shows the internal electronics of the device, where the primary components are distributed across two printed circuit boards (PCBs). The first board (PCB1), located at the top, contains the Hall effect sensor, which measures the knee joint angle based on the relative motion of the magnet attached to the lever arm during flexion and extension.

The second board (PCB2) integrates the system's core electronics, including the microcontroller, voltage regulator, and EMG acquisition modules.

5) Base Compartment: Finally, Fig. 1(e) presents the base compartment of the device, which houses the LiPo battery in a dedicated space. Although the protective cover is not shown for visualization purposes, this section also includes Velcro attachment holes to ensure the device remains stable during use.

C. Control Design

This section describes the control design of the proposed device. Fig. 2 presents the system's operational logic for data acquisition, reception, and transmission.

The process begins with a connection verification stage, during which the microcontroller checks the sensors' proper operation. If a fault is detected, the system notifies the user and halts the test. Next, the battery level is evaluated using an integrated voltmeter. If the charge is below the required threshold, the device powers off and must be recharged before continuing. Once these conditions are satisfied, the system enters a standby mode, awaiting the user's start command.

Upon test initiation, data acquisition from the EMG sensors and the angular measurement sensor is enabled. Time vectors are generated to support real-time visualization and subsequent analysis. The acquisition process continues until a stop command is received, after which the recorded data are transmitted in matrix format for further processing. The hardware interconnections are further detailed in Fig. 3. The system is powered by a 3.3 V LiPo battery controlled by a switch. A PowerBoost 1000C module regulates the supply voltage and powers the Beetle BLE microcontroller, the analog EMG sensors, and the AS5048B Hall-effect sensor. A digital voltmeter continuously monitors the battery level; when it drops below 10%, data acquisition is stopped, and the system prompts the user to recharge the device via USB.

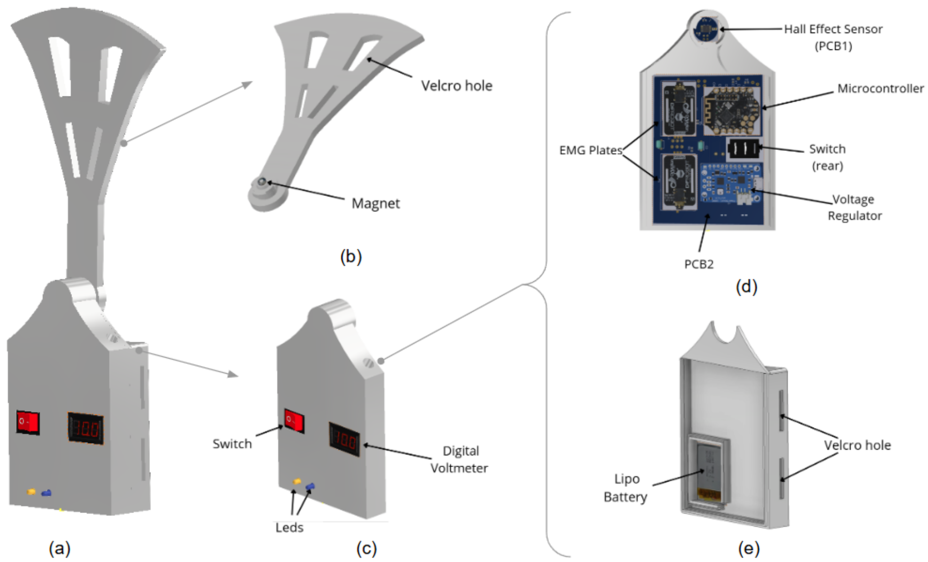

Fig. 1. Mechanical and electronic design of the proposed device: (a) assembled device, (b) lever arm, (c) front view, (d) internal electronics, (e) base compartment.

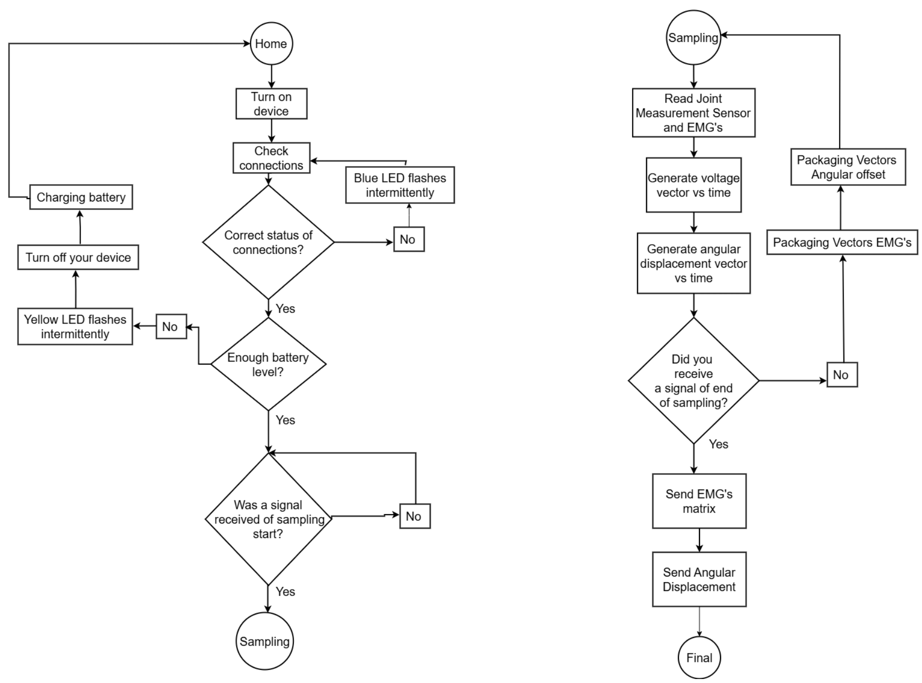

Fig. 2. System operational block diagram.

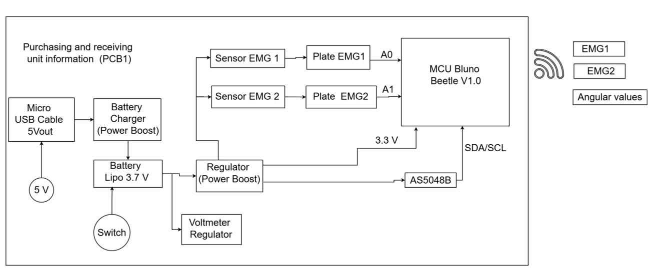

Fig. 3. Hardware connection block diagram.

The microcontroller verifies the integrity of the EMG signals acquired through analog pins (A0 and A1) and the I^2C communication with the AS5048B sensor. In case of communication or signal faults, the system returns to standby mode until a new command is received. Subsequently, an individual calibration process is performed, including zero-angle adjustment at the neutral position and offset compensation of the EMG sensors under resting conditions. Knee joint angle during gait is measured using the AS5048B Hall-effect sensor, which detects magnetic field variations generated by the rotation of a lever arm attached to the knee joint during movement. These magnetic field variations are provided as 14-bit digital output values, which are subsequently converted into angular displacement as in (1).

$$\theta = \frac{\text{Sensor output}}{16384} \times 360^\circ \tag{1}$$

In this equation, θ represents the calculated knee joint angle, sensor output corresponds to the AS5048B digital value, and 16384 represents its full 14-bit resolution over 360° [6], [8]. After calibration, the system performs synchronized acquisition, sampling EMG signals at 1000 Hz while satisfying the Nyquist criterion, while angular measurements are acquired at 100–200 Hz.

III. Results

This section presents results from signals acquired with the proposed device, including electromyographic (EMG) activity and knee joint kinematics throughout the gait cycle.

A. Electromyographic Signals

In this section, several processing iterations were performed, and the data were segmented to identify and isolate phases of increased muscular activity, particularly distinguishing between the swing and stance phases of the gait cycle.

Based on the literature [19], the rectus femoris and vastus lateralis are commonly associated with knee extension, whereas the biceps femoris and semitendinosus primarily contribute to knee flexion. Due to hardware limitations, only two muscles were monitored; therefore, one representative muscle was selected for each movement: the rectus femoris (extension) and the biceps femoris (flexion). These muscles were chosen because of their ease of electrode placement and their common use in standardized EMG acquisition protocols, particularly according to SENIAM's guidelines.

As shown in Fig. 4, the analog EMG sensors are positioned on the selected muscles, with electrode placement performed according to SENIAM's guidelines to ensure standardized and reliable muscle activity acquisition [20].

A gait test covering three cycles was conducted under controlled conditions. The acquired EMG signals were segmented into two main phases of the gait cycle: stance (initial phase) and swing (final phase), enabling a clearer interpretation of muscle activation patterns.

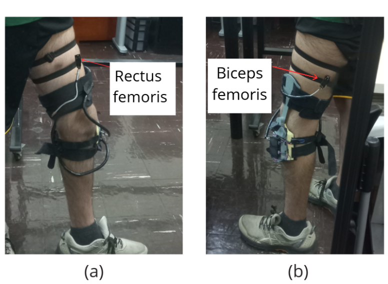

Fig. 4. Positioning of EMG Sensors on Muscles of Interest

1) Biceps Femoris: As shown in Fig. 5, the biceps femoris exhibits minimal electrical activity during the stance phase, indicating limited involvement in this stage of the gait cycle. In contrast, during the swing phase, a significant increase in signal amplitude is observed, reaching peak values of approximately 60 µV, as illustrated in Fig. 6. This behavior is consistent with the functional role of the biceps femoris in knee flexion and limb deceleration during the swing phase.

2) Rectus Femoris: Fig. 7 presents the EMG signal of the rectus femoris during the stance phase, where a high level of activation is observed, with peak values close to 120 µV. This reflects its role in knee stabilization and extension during weight-bearing. During the swing phase (Fig. 8), the signal amplitude decreases considerably, indicating reduced activation. This behavior aligns with the diminished contribution of the rectus femoris once the limb transitions into forward motion. Overall, the observed EMG patterns are consistent with the expected physiological activation of these muscles during gait, supporting the reliability of the proposed measurement system.

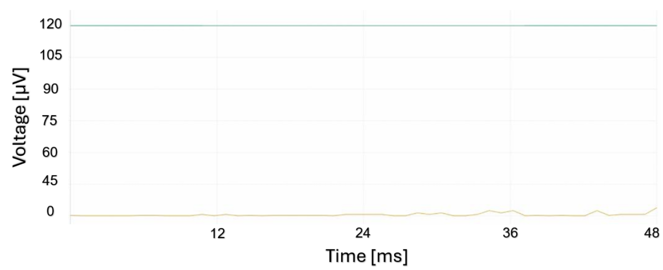

Fig. 5. EMG signal of the biceps femoris during the stance phase (µV vs. time in ms).

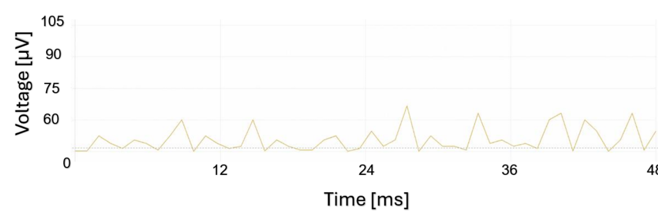

Fig. 6. EMG signal of the biceps femoris during the swing phase (µV vs. time in ms).

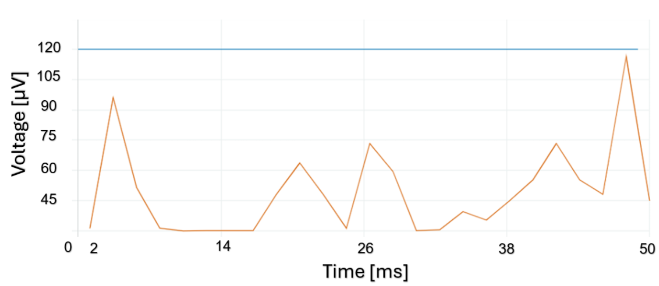

Fig. 7. EMG signal of the rectus femoris during the stance phase (µV vs. time in ms).

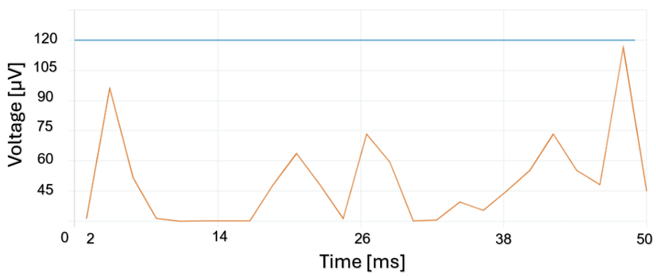

Fig. 8. EMG signal of the rectus femoris during the swing phase (µV vs. time in ms).

B. Joint Range of Motion

To evaluate the dynamic performance of the prototype, a continuous walking test (Fig. 9) consisting of three complete gait cycles was initially conducted. The angular displacement ranges approximately from 0° to 65°, capturing both knee flexion and extension phases. This allows verification of a repetitive angular pattern consistent with normal knee kinematics.

Based on this preliminary assessment, seven additional iterations were performed to obtain a statistical analysis of the joint range of motion using precision and repeatability metrics. For this purpose, in Table I, performance metrics such as Mean Absolute Error (MAE) were calculated to quantify the angular accuracy of the system relative to the conventional goniometer. At the same time, Mean Absolute Deviation (MAD) was used to assess the variability and internal consistency of the acquired measurements. The statistical analysis yielded a MAD of 3.09° and an MAE of 1.67°, indicating good precision and acceptable variability of the device. The MAE must be below 2° to be considered optimal [21], [22], and the MAD must be below 3.5° for optimal performance [23].

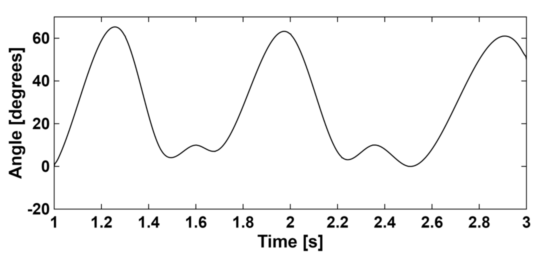

Fig. 9. Knee joint angle (degrees) as a function of time (s).

TABLE I
STATISTICAL ANALYSIS OF KNEE JOINT ANGULAR MEASUREMENTS

Iterations	MAD (°)	MAE (°)
1	2.88	1.42
2	3.10	1.56
3	2.97	1.44
4	3.12	1.65
5	3.01	1.88
6	3.15	1.96
7	3.43	1.77
Overall	**3.09**	**1.67**

Also, the obtained range of motion falls within the expected values reported in gait analysis studies [6], [24], indicating that the system can accurately track knee kinematics. Additionally, the smooth signal variation suggests stable sensor performance and proper synchronization with the EMG acquisition.

IV. DISCUSSION

Given that this is a recent prototype, data collection across multiple participants was not feasible. However, several trials were repeated to ensure consistency and improve the reliability of the statistical results.

A. Electromyographic Signals

The analysis of the biceps femoris muscle reveals low electrical activity during the stance phase, with near-baseline values, and a moderate increase during the swing phase, reaching amplitudes of 30–40 µV. This behavior is consistent with previous findings reported in [19], where the biceps femoris shows limited activation during the initial phases of gait and increased involvement during the terminal phases, primarily associated with knee flexion and limb deceleration.

In contrast, the rectus femoris exhibits an inverse activation pattern. During the stance phase, its electrical activity is significantly higher, with peak values approaching 120 µV, particularly toward the end of the analyzed segment. As the gait cycle progresses into the swing phase, the activity decreases substantially to near-minimal levels. This trend aligns with established studies [12], [19], which describe the rectus femoris as a primary contributor to knee extension, with peak activation occurring during weight-bearing and reduced involvement during limb advancement.

Overall, the EMG patterns obtained in this study are consistent with the physiological roles of the analyzed muscles, supporting the proposed system's ability to capture meaningful neuromuscular activation during gait. Although the experimental validation was conducted on three gait cycles under controlled conditions, the results show the system's capability to capture physiologically consistent neuromuscular and kinematic patterns, supporting its feasibility for future extended studies.

Due to current hardware limitations, this initial prototype focused on monitoring two key muscles for gait analysis: the rectus femoris and biceps femoris. However, future iterations could incorporate additional muscle groups, such as the semitendinosus and vastus lateralis, using additional analog EMG sensors, though this may require a higher-capacity microcontroller to maintain real-time acquisition.

B. Joint Range of Motion

The measured knee joint range of motion varies between 0° and 65°, where 0° corresponds to full knee extension. These values are in close agreement with those reported using conventional goniometric methods [6], [24], [25].

The Hall-effect sensor is robust for clinical use due to its non-contact design, which reduces sensitivity to electrode displacement and sweat. Its direct angle measurement minimizes cumulative errors compared to inertial systems, although proper fixation remains essential. It also demonstrates good stability for long-duration gait monitoring.

Furthermore, the system demonstrates a measurement error of 1.67°, and standard deviation of 3.09°, which is below the commonly accepted threshold of 2.00° and 3.50° for each measure. This level of accuracy supports the reliability and repeatability of the proposed device for capturing joint angle variations. Additionally, the consistency and smoothness of the recorded angular signal suggest stable sensor performance and appropriate system calibration. The system emphasizes embedded acquisition, sensor synchronization, and real-time signal processing, aligning with control and instrumentation applications. This result was validated by comparison with a conventional goniometer. However, no comparison was performed against an optical motion capture system, as physical verification of such measurements was not feasible within the scope of this study.

V. CONCLUSIONS

The results show that the proposed device successfully meets the main objective of this study, providing a portable, accurate, accessible, and user-friendly system for gait analysis through integrated mechanical, electronic, and control design. The electromyographic measurements exhibit activation patterns consistent with the physiological behavior of the analyzed muscles. In particular, the biceps femoris shows higher activity during the swing phase. In contrast, the rectus femoris reaches peak activation during the stance phase, in agreement with reported findings in the literature.

In addition, the knee joint angle measurements were reliable, with a mean absolute error of 1.67° compared with conventional goniometry and a mean absolute deviation of 3.09°. This value is below the commonly accepted threshold of 2° and 3.5° respectively, confirming the accuracy and robustness of the proposed system. Overall, the results validate the device's ability to simultaneously capture neuromuscular activity and joint kinematics, supporting its potential application in gait assessment.

The device incorporated design considerations to reduce the impact of electrode displacement, sweat accumulation, motion artifacts, and prolonged use through standardized SENIAM-based electrode placement, as well as integrated amplification and signal filtering. Additionally, its lightweight design and short-duration acquisition protocol improved stability during testing; however, future versions should further optimize electrode fixation, sweat resistance, and adaptive filtering for extended clinical applications.

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