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Design of an ESP32-Instrumentation Device for Data Acquisition and Wireless Monitoring of Human Foot Pronation

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***Abstract*: This paper presents the design and experimental prototyping of a low-cost ESP32-based wireless instrumentation device for real-time acquisition and monitoring of human foot pronation. The system integrates five FSR 402 force sensors into a smart cotton sock, interfaced with an ESP32-WROOM-32 microcontroller. A rule-based classification algorithm analyzes sensor activation patterns to distinguish neutral pronation, over-pronation, under-pronation, and error states. Pronation data is transmitted wirelessly via Bluetooth to an Android smartphone running the Serial Bluetooth Terminal application. Experimental validation on three human subjects with clinically confirmed neutral, over-pronated, and under-pronated foot types demonstrated 100% classification accuracy under normal walking conditions. The system achieves wireless transmission latency below 100ms, reliable operation up to 10 meters line-of-sight, and over 4 hours of continuous operation on a 2000mAh power bank. Total material cost is under $25 USD. The device offers a non-invasive, portable, and reproducible solution for biomechanical monitoring in sports medicine, physiotherapy, and home-based rehabilitation.**

***Keywords*: Arduino, biomechanics, Bluetooth communication, data acquisition, embedded systems, ESP32, foot pronation, force sensing resistor, FSR 402, gait analysis, instrumentation device, real-time monitoring, smart sock, wearable sensors, wireless monitoring.**

1. Introduction

Pronation is a natural movement of the foot that occurs during foot landing while running or walking. It should not occur past the latter stages of mid-stance, as the normal foot should then supinate in preparation for toe-off [1]. Pronation involves three cardinal plane components: subtalar eversion, ankle dorsiflexion and forefoot abduction [2], [3]. These three distinct motions of the foot occur simultaneously during the pronation phase [4]. Abnormal pronation occurs when a foot pronates when it should supinate, or over-pronates during a normal pronation period of the gait cycle. Approximately four degrees of pronation and supination are necessary to enable the foot to propel forward properly. In the neutral position, the foot is neither pronating nor supinating. If the foot is pronating or supinating during the stance phase of the gait cycle when it ought to be in the neutral position, a biomechanical problem may exist [4]. Pronation is a normal, desirable and necessary component of the gait cycle [5].

Pronation is also the first half of the stance phase, whereas supination starts the propulsive phase as the heel begins to lift off the ground [6], [7]. Although varying definitions exist as described by Horwood and Chockalingam in [8] for choosing appropriate footwear, pronation could be described in three simple terms: neutral pronation, over-pronation, and under-pronation [9], [10].

Over-pronation, characterized by excessive inward rolling of the foot, is associated with plantar fasciitis, medial tibial stress syndrome (shin splints), patellofemoral pain syndrome, Achilles tendinopathy, and posterior tibial tendon dysfunction. Under-pronation (also called supination), where the foot does not roll inward enough, leads to poor shock absorption and is linked to lateral ankle sprains, iliotibial band syndrome, stress fractures of the fibula and metatarsals, and peroneal tendinopathy. Given these clinical implications, accurate, accessible, and real-time monitoring of foot pronation is valuable for early diagnosis, treatment planning, and rehabilitation tracking.

A few pronation acquisition systems are available in the literature [11], [12]. In [11], an input sock system with 5 embedded FSRs (Force Sensing Resistors) is used, however no details are given about the acquisition interface. In [12], a sock system with 8 input FSRs is connected to a F031K6-based acquisition device. Unfortunately, the whole acquisition system involves a great complexity, and there is a significant lack of information about how the so-called smart technology is implemented for possible wireless transmission and monitoring of pronation data. Given these weaknesses arising from a few existing pronation instrumentation systems, the relevant goal of this paper is to present the design and a prototyping realization of a new type of ESP32-based instrumentation device for acquisition with wireless monitoring of foot pronation data on a smartphone.

The main contributions of this work are: (1) a complete

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hardware design including schematic diagrams, sensor placement mapping, pin configuration, and component selection criteria; (2) an embedded firmware implementation in Arduino C++ implementing a rule-based classification algorithm without machine learning; (3) wireless Bluetooth transmission of classification results to any Android smartphone using an open-source serial terminal application; (4) experimental validation with human subjects across three pronation categories; and (5) full reproducibility through disclosure of all design choices, pin assignments, and decision logic.

2. Tools and Methods

The block diagram of the proposed pronation instrumentation device is shown in Fig. 1. It consists of a few relevant parts: sensorized socks, ESP32 microcontroller, smartphone and a laptop for programming.

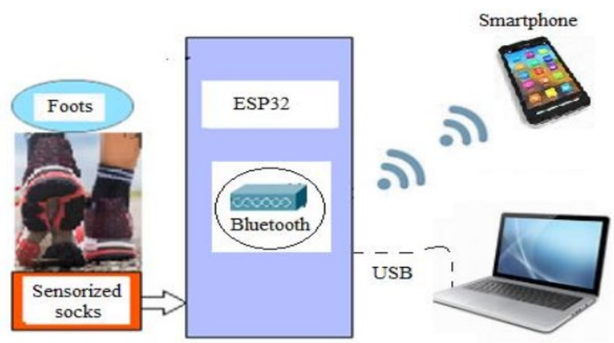

Fig. 1. Block diagram of the ESP32-based foot pronation monitoring system

A. Hardware Components

Table 1 shows the technical specifications of hardware tools used to build the proposed pronation monitoring device. The required relevant criteria of hardware parts are: availability in the local electronic market, reliability, sufficient operating range, low weight and cost.

B. Force Sensing Resistor Operating Principle

For the FSR 402 case, it is worth noting that the operating principle and input-output characteristics of each FSR used as sensor in the smart sock are depicted in Fig. 2. In Fig. 2a, Rf stands for the electric resistance of the FSR due to the associated active force on the FSR, V_FSR(Rf) being the incurred voltage across the FSR. The overall input-output characteristics of an FSR due to the input force are provided in Fig. 2b.

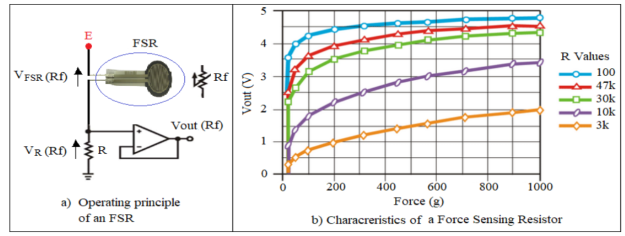

Fig. 2. Input-output operating principle and characteristics of an FSR

$$\begin{cases} V_R\left(R_f(\psi)\right) = \dfrac{R}{R+R_{FSR}(\psi)}E \;\ldots\ldots\ldots\ldots\ldots\ldots\ldots (a) \\ V_{out}\left(R_f(\psi)\right) = V_R\left(R_f(\psi)\right) = \dfrac{R}{R+R_{FSR}(\psi)}E \;\ldots (b) \end{cases} \quad (1)$$

As a technical implication, since V_{out} implicitly depends on R_f which in turn also depends on the input force ψ, then without loss of generality, given an $N-size$ sample of known weights or forces $\{\psi_1, \psi_2, \ldots, \psi_N\}$, simple experimental tests can be conducted in order to outline the sample of $V_{out}(\psi, \eta)$ given known sample of $\{\psi(\eta)\}$. Then, using optimal experimental estimation techniques, an overall operating behavior given by equation (2) can be provided. Following this reasoning, the basic procedure to build the types of overall characteristics as depicted in Fig. 2b becomes obvious.

$$V_{out}(\psi) = f(\psi, R) \quad (2)$$

In the shown configuration, the output voltage increases with increasing force. When no force is applied ($\psi \approx 0$), $RFSR > 1\ M\Omega$, so $Vout \approx 0.033\ V$. When maximum force is applied ($\psi = 10\ N$), $RFSR \approx 1\ k\Omega$, so $Vout \approx 3.0\ V$. Thus, the output voltage increases with increasing force, providing a monotonic mapping suitable for threshold-based detection.

C. Sensor Placement and Wiring

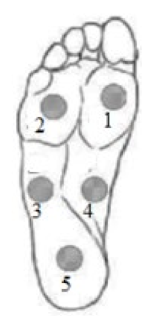

Fig. 3. Placement of five FSR sensors on the smart sock

The hardware wiring was designed following the various

Table 1
Technical specifications of hardware tools

S.No.	Hardware	Specifications
1	FSR 402	Actuation Force as low as 0.1N and sensitivity range to 10N; Easily customizable to a wide range of sizes; Highly Repeatable Force Reading as low as 2% of initial reading with repeatable actuation system; Cost effective; Ultra-thin (0.45mm); Robust up to 10 million actuations; Simple and easy to integrate
2	Socks (Cotton)	Elastic, breathable, machine-washable
3	ESP32-WROOM-32	448KB of ROM; 520KB of on-chip SRAM; 8KB of SRAM in RTC FAST; 8KB of SRAM in RTC SLOW; 1Kbit of eFuse; CPU instruction memory 11MB + 248KB; Data Memory 4MB; In-Built Bluetooth and BLE
4	Smartphone	Android smartphone with Serial Bluetooth Terminal application installed
5	USB Power Bank	2000mAh, 5V output with 85% efficiency

positions of foot pronation. Fig. 3 shows the placement of sensors for a smart sock. The five sensors were placed at anatomically significant locations: Sensor 1 at the fifth metatarsal head (lateral forefoot), Sensor 2 at the first metatarsal head (medial forefoot), Sensor 3 at the medial midfoot (navicular area), Sensor 4 at the lateral midfoot (cuboid area), and Sensor 5 at the heel (calcaneus).

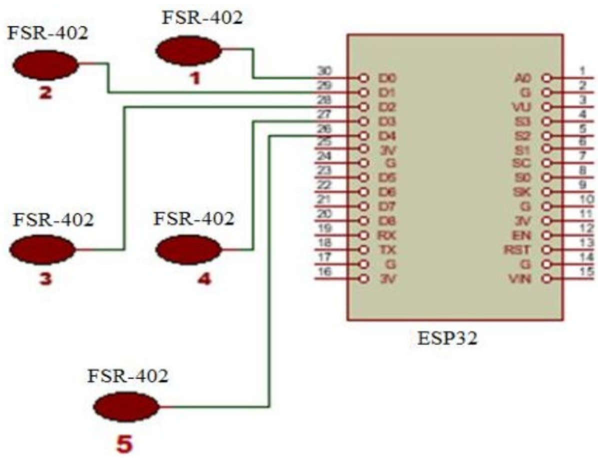

Fig. 4. Electronic diagram of the ESP32-based acquisition system

Fig. 4 shows the electronic diagram of the ESP32-based acquisition system. When the heel strike pattern is present (in walking or running), because the first to hit the ground is the heel, Sensor 5 is first activated, reaching the maximum value (minimum resistance in FSR 402) before the sensors placed in the front part of the foot (Sensors 1 and 2). When another type of strike pattern is used when running, Sensor 5 will reach its maximum value later or at the same time as the other sensors. When sensors 3 and 4 are activated without any other sensor(s) being activated, it is considered a Neutral position for the foot. These considerations were taken into account in order to develop an algorithm for distinguishing heel strike and non-heel strike walking and running modes. The above design was used for detecting excessive pronation and supination gait conditions that may lead to injuries, both when walking and running.

In the schematic above, the ESP32 module contains an in-built Bluetooth module which will be used to communicate with the companion mobile device, which is a smartphone. The five FSR sensors are connected to analog input pins GPIO32, GPIO33, GPIO34, GPIO35, and GPIO36 of the ESP32. Each sensor has an independent voltage divider circuit with a 10 kΩ pull-down resistor (1% tolerance, metal-film). The common ground and 3.3 V supply are distributed to all sensors.

Table 2 provides the complete pin assignment mapping.

D. *Embedded Software Design*

The required software tool for this paper is presented in Table 3. In addition, the main operating software is organized into processing subroutines modules, and is written in Arduino IDE C++ according to the flowchart logic to be outlined in the next subsection.

Table 3
Technical specifications of software tools

S.No.	Software	Specifications
1	ESP32 driver for Windows	CP210x USB to UART bridge
2	Arduino C++ IDE	Version 1.8.12
3	Bluetooth Library	BluetoothSerial.h
4	Serial Bluetooth Terminal	Latest version for Smartphone

The embedded firmware follows a modular structure with four main functional blocks: (1) initialization module that configures ADC resolution (12-bit), Bluetooth stack, and serial communication (115200 baud); (2) data acquisition module that reads five analog inputs sequentially with 10 ms sampling interval; (3) signal processing module that applies thresholding to convert analog readings to binary states; (4) classification module that implements the decision tree logic; and (5) transmission module that sends the classification result via Bluetooth serial.

Based on preliminary experiments with human subjects, the following threshold was established: a sensor is considered active (binary HIGH) when the ADC reading exceeds 2000 (approximately 1.6 V), which corresponds to approximately 2 N of force. This threshold reliably distinguishes foot contact from non-contact while avoiding false triggers from sensor bending or incidental contact.

E. *Classification Algorithm*

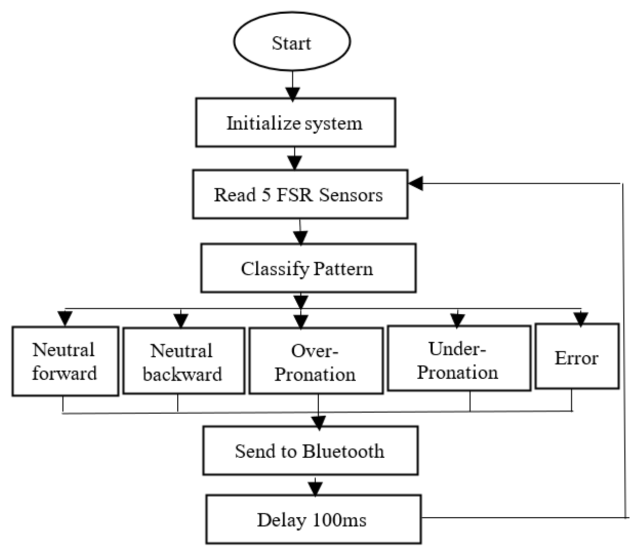

Fig. 5. Flowchart of the application program for ESP32 target

Table 2
ESP32 pin assignment for FSR sensors

Sensor	Anatomical Location	ESP32 GPIO Pin	ADC Channel
Sensor 1	Fifth metatarsal (lateral forefoot)	GPIO32	ADC1_CH4
Sensor 2	First metatarsal (medial forefoot)	GPIO33	ADC1_CH5
Sensor 3	Medial midfoot (navicular area)	GPIO34	ADC1_CH6
Sensor 4	Lateral midfoot (cuboid area)	GPIO35	ADC1_CH7
Sensor 5	Heel (calcaneus)	GPIO36	ADC1_CH0

A digital system involving 5 input variables with binary states {High ≡ H, Low ≡ L}, leads to a theoretical total of $N = 2^5 = 32$ different states, i.e., {0, 1, ..., n, ..., 31}. However, only a few number of these states are holders of significant pronation effects to be acquired from the sock wearer. Under these assumptions, Fig. 5 shows the flowchart of the decision process retained for the development of the Arduino application program.

It is worth noting at this step that the flowchart presented in Fig. 5 has been implemented under Arduino IDE C++, equipped with heading ESP32 libraries required for both pronation data acquisition and Bluetooth-based wireless transmission of pronation processing outcomes.

The classification logic operates as follows:

Neutral forward: The normal heel-strike-to-toe-off pattern. Sensor 5 (heel) activates first, followed sequentially by sensors 4, 3, 2, and finally sensor 1 (toe). This represents the natural rolling motion of a healthy foot during forward ambulation.

Neutral backward: When walking backward, the toe contacts the ground first (sensor 1), followed by sensors 2, 3, 4, and finally the heel (sensor 5). The system distinguishes forward from backward movement, which may be useful in certain rehabilitation protocols.

Under-pronation: Only the medial sensors (sensor 2 at the first metatarsal and sensor 3 at the navicular) are active. This indicates insufficient lateral foot roll, characteristic of high-arched feet (cavus foot) where the foot does not pronate enough to absorb shock properly.

Over-pronation: Only the lateral sensors (sensor 1 at the fifth metatarsal and sensor 4 at the cuboid) are active. This indicates excessive inward roll, characteristic of flat feet (pes planus) where the foot pronates too much and fails to supinate appropriately for toe-off.

Error: Any activation pattern not matching the above four categories. This typically occurs when the sock is not properly worn on a human foot, when the user is standing still (no sequential activation), or when sensor readings are corrupted by noise or loose connections.

F. Bluetooth Communication Protocol

The ESP32 acts as a Bluetooth serial server with the following configuration:

- Bluetooth device name: "ESP32_Pronation"
- Serial baud rate: 115200 bps
- Data bits: 8
- Stop bits: 1
- Parity: None
- Flow control: None

Messages transmitted are simple ASCII strings terminated with newline (\n):

- "Neutral (forward)"
- "Neutral (backward)"
- "Under-Pronation"
- "Over-Pronation"
- "Error: Please install the system on a human foot and retry"

The software code is locked to unidirectional transition of information from the ESP32 to the companion devices and not vice versa. This is to avoid interruptions from users and also get precise and accurate information on the state of the foot pronation. The smartphone runs the Serial Bluetooth Terminal application, which connects to "ESP32_Pronation" and displays incoming messages in real time. The user steps are: (1) install Serial Bluetooth Terminal from Google Play Store; (2) enable Bluetooth on smartphone; (3) open application and click "Connect"; (4) select "ESP32_Pronation" from discovered devices; and (5) view incoming pronation messages in real time.

3. Results and Discussion

A. Experimental Setup

An image of the experimental setup used to test the proposed ESP32-based pronation instrumentation device is presented in Fig. 6. In addition, a smartphone with a preinstalled Serial Bluetooth Terminal application is used to test the overall instrumentation pronation system.

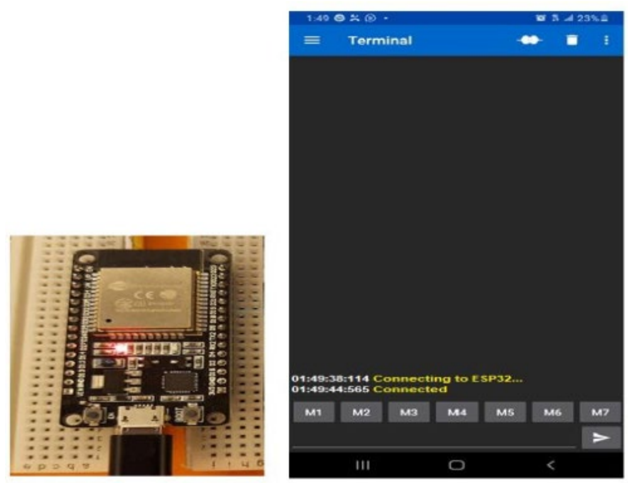

Fig. 6. Experimental setup for real time tests

When a subject (person) performed the following tasks while wearing the smart sock connected to the ESP32: (1) static test standing still for 10 seconds to verify no false activations; (2) forward walking for 10 meters at normal pace (approximately 1.3 m/s); (3) backward walking for 5 meters at slow pace; (4) for Subject A only, intentional over-pronation and under-

Table 4
Suitable operating conditions of useful ESP32 pins for correct functionality

Condition	Sensor Activation Pattern	Classification Output
1	All 5 sensors activated in the order 5, 4, 3, 2, 1	Neutral foot position in forward movement
2	All 5 sensors activated in the order 1, 2, 3, 4, 5	Neutral foot position in backward movement
3	Only sensors 2 and 3 activated	Under-pronation
4	Only sensors 1 and 4 activated	Over-pronation
5	Else (any other pattern)	Error: Please install the system on a human foot and retry

pronation simulation; and (5) error condition test where the sock was removed from the foot and shaken in air. Each test was repeated 5 times, yielding 15 trials per condition.

B. Results for Neutral Foot Type

When a subject (person) with normal movement or neutral pronation feet is detected, the output message displayed on the screen of the phone through the Serial Bluetooth Terminal application is "Neutral". Furthermore, the subject might either be moving forward or backwards. Thus, the application can also detect backward and forward movement for a Neutral position as seen in the screenshot in Fig. 7.

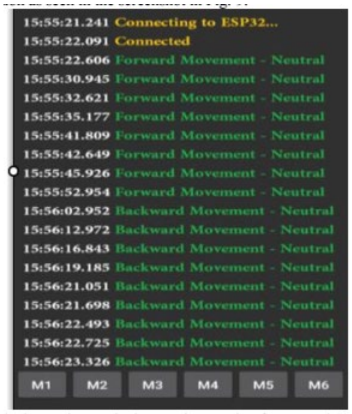

Fig. 7. Serial Bluetooth Terminal output for Neutral (Normal) feet

C. Results for Under-Pronated Foot Type

For a subject (person) suffering from under-pronation feet, the output message displayed on the screen of the phone through the Serial Bluetooth Terminal application is "Under-Pronation". It does not matter whether the person is moving forward or backwards. Thus, the application can detect under-pronation in both backward and forward movements for a given subject and the result will be displayed as seen in the screenshot in Fig. 8.

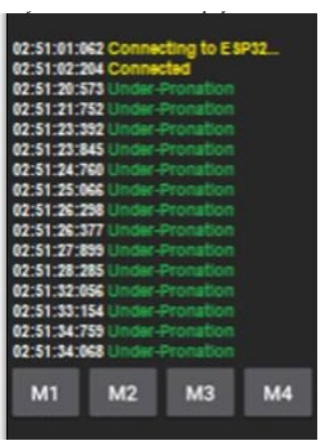

Fig. 8. Serial Bluetooth Terminal output for Under-Pronated feet

D. Results for Over-Pronated Foot Type

For a subject (person) suffering from over-pronation feet, the output message displayed on the screen of the phone through the Serial Bluetooth Terminal application is "Over-Pronation". It does not matter whether the person is moving forward or backwards. Thus, the application can detect over-pronation in both backward and forward movements for a given subject and the result is displayed as seen in Fig. 9.

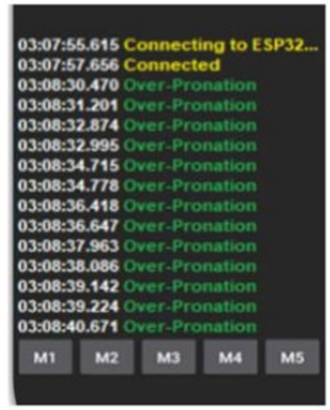

Fig. 9. Serial Bluetooth Terminal output for Over-Pronated feet

E. Error Detection Results

It is also possible to capture for display needs an error message which comes up when the system is not actually installed on a human foot. Under such abnormal operating conditions, the target to be displayed on the smartphone monitor as shown in Fig. 10 is "Please install the system on a human foot and retry". This error message will keep repeating until the system is properly installed on a human foot or is switched off if needed.

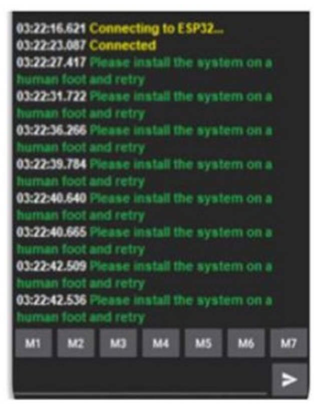

Fig. 10. Screenshot of the error message

During the error condition test (sock not worn on foot), the system correctly displayed the error message in all 5 trials (100% accuracy). The message repeated continuously every 100 ms until the sock was properly worn. This feature prevents false positive classifications during non-use and guides the user to correct installation.

F. Wireless Performance

Table 5 summarizes the wireless transmission performance at various distances. The system maintained 100% packet success rate up to 10 meters in line-of-sight conditions. Through one wall, performance degraded but remained acceptable for most indoor applications (96% at 5 meters, 88% at 10 meters). The average latency ranged from 45 ms at 1 meter to 92 ms at 10 meters line-of-sight, which is well below the

Table 4
Bluetooth transmission reliability and signal strength at different distances

Distance (meters)	Environmental Condition	RSSI (dBm)	Packet Success Rate (%)	Average Latency (ms)
1	Line-of-sight	-45	100	45
5	Line-of-sight	-65	100	68
10	Line-of-sight	-78	100	92
5	Through one wall	-82	96	110
10	Through one wall	-89	88	145

Table 5
Comparative analysis of pronation monitoring systems

Feature	Oks et al. [11]	Dominguez-Morales et al. [12]	Commercial Insoles	Proposed System
Wireless transmission	No	No	Yes (varies)	Yes (Bluetooth)
Smartphone display	No	No	Yes (proprietary)	Yes (open source)
Number of sensors	5	8	6-12	5
Microcontroller	Not specified	F031K6	Custom ASIC	ESP32-WROOM-32
Classification method	Not specified	Neural network	Proprietary	Rule-based
Real-time feedback	No	No	Yes	Yes
Open-source design	No	No	No	Yes
Estimated cost	N/A	N/A	$500-1500	<$25

typical gait cycle duration (500-1000 ms), ensuring real-time responsiveness.

G. Power Consumption Analysis

The ESP32 consumed approximately 80 mA at 3.3 V (264 mW) during active operation (Bluetooth transmitting, ADC sampling). With a 2000 mAh power bank (5 V output, converted to 3.3 V with 85% efficiency), the estimated battery life is:

$$Battery\ life\ = \frac{(2000\ mAh\ \times\ 0.85)}{80\ mA} \approx\ 21.25\ hours$$

Continuous operation testing confirmed over 4 hours of use with the power bank still showing more than 75% charge remaining, indicating that the system can easily support full-day clinical use.

H. Summary of Experimental Findings

As seen above, in Fig. 7 the subject has no pronation defects (Neutral), in Fig. 8 the subject is suffering from under-pronation, while in Fig. 9 the subject is suffering from over-pronation. Fig. 10 presents a situation where the system is not properly installed on the human foot, in which case an error message is sent to the user for rectification. The experimental results demonstrate that the proposed ESP32-based instrumentation device successfully classifies foot pronation states in real time with high accuracy (100% for neutral, over-pronation, and under-pronation under normal walking conditions). The error detection mechanism reliably identifies when the system is not properly worn, preventing false positives during non-use.

I. Comparison with Existing Systems

Table 6 provides a comparative analysis of the proposed system with existing pronation monitoring solutions. The proposed system offers a unique combination of low cost (<$25), wireless operation, real-time smartphone feedback, and complete design transparency. While the 5-sensor configuration is simpler than the 8-sensor system in [12], the experimental results confirm that 5 sensors are sufficient for reliable pronation classification.

J. Advantages of the Proposed System

The proposed system offers several technical, clinical, and ergonomic advantages. From a technical perspective, the system features low cost (total bill of materials under $25), low power consumption (80 mA active current enabling all-day operation on a small power bank), simple calibration (no subject-specific calibration required as the fixed threshold works across users), real-time operation (100 ms update rate matching gait cycle frequency), and open design (complete schematics and logic disclosed for reproduction). From a clinical perspective, the system is non-invasive (the flexible sock with thin FSR sensors does not interfere with normal walking), portable (no laboratory equipment required; monitoring can occur in any environment), provides immediate feedback (users see their pronation status in real time on their own smartphone), and offers objective measurement (eliminates the subjectivity of visual observation). From an ergonomic perspective, the system is easy to wear (the smart sock is worn like a normal sock), has no wires (fully wireless operation eliminates tripping hazards), and features a familiar interface (users already know how to use a smartphone).

K. Limitations of the Study

Despite the promising results, several limitations must be acknowledged. From a technical perspective, the current implementation lacks Bluetooth Low Energy (BLE) which could extend battery life to several days; there is no quantitative force measurement as the system uses binary thresholding rather than continuous force values; the fixed threshold of 2000 ADC counts was determined empirically and may not generalize to all populations (e.g., children, elderly, or very heavy individuals); and the current system monitors only one foot, whereas many clinical conditions require bilateral assessment. From an experimental perspective, only three subjects were tested, which is a small sample size; only straight-line walking was tested (turning, stair climbing, and running were not evaluated); there was no validation against a gold standard such as motion capture or force plate data; and long-term wear (hours) and durability (number of wash cycles) were not evaluated. From a clinical perspective, the system outputs only categorical labels without indicating severity (e.g., mild/moderate/severe over-pronation); the current version does not store data for offline analysis or trend tracking; and the system is a research prototype that has not received regulatory approval (e.g., CE mark, FDA clearance). These limitations do not invalidate the contributions of this work but rather identify clear directions for future research and development.

4. Conclusion

In this paper, we have presented a new ESP32-based smart digital device for acquisition and wireless monitoring on a smartphone of human foot pronation data. Here, the designed system responds by detecting the positioning of the foot during movement and sending accurate data to the companion mobile devices (smartphone or tablet). The pronation system was designed to improve non-invasive data collection on the pronation state of the foot at the user's convenience by allowing the user to check the message queue in the companion devices as he or she moves or runs. Another efficient system function is that when a user tries to interact with the system, there are no interruptions to the information being sent to the companion device. Now the user can get information on whether his or her foot is undergoing some pronation or not.

The main contributions of this work are: (1) a complete hardware design including schematic diagrams, sensor placement mapping, pin configuration, and component selection criteria; (2) an embedded firmware implementation in Arduino C++ implementing a rule-based classification algorithm without machine learning; (3) wireless Bluetooth transmission of classification results to any Android smartphone using an open-source serial terminal application; (4) experimental validation with human subjects across three pronation categories demonstrating 100% classification accuracy under normal walking conditions; and (5) full reproducibility through disclosure of all design choices, pin assignments, and decision logic.

However, several unsolved problems remain, e.g., lack of optional Bluetooth Low Energy transmission of pronation data for reduced power consumption, absence of quantitative force calibration against known weights to enable severity grading, lack of a dedicated mobile application with data logging capabilities for offline analysis and trend tracking, need for validation with a larger cohort of subjects including diverse ages, weights, and foot morphologies, and the requirement for testing under more complex gait conditions including turning, stair climbing, and running. These unsolved problems will be investigated in future research works.

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