A Low-Cost Flex Sensor Gait Phase Gating Method for Powered Ankle Prostheses

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Citation | Ghafoor. A, Awan. M. I, Hassan. S, S, "A Low-Cost Flex Sensor Gait Phase Gating Method for Powered Ankle Prostheses", IJIST, Vol. 8 Issue. 3 pp 1366-1378, June 2026
DOI: https://doi.org/10.33411/IJIST/1931
Received| May 06, 2026 **Revised|** June 08, 2026 **Accepted|** June 12, 2026 **Published|** June 21, 2026.

Powered ankle prostheses restore the active push-off that passive prosthetic feet cannot provide, yet their cost places them beyond reach for most lower-limb amputees in low- and middle-income countries. This work presents a low-cost knee motion sensing and gait phase gating method for a powered ankle prosthesis using a single resistive flex sensor costing about one thousand seven hundred Pakistani rupees, roughly six United States dollars. The flex sensor is mounted across the knee region of the prosthetic socket and sampled at one hundred hertz on an ESP32-S3 microcontroller. A two-point calibration maps the sensor voltage to knee flexion angle, and a gait phase gate uses the calibrated angle to suppress electromyography classifier outputs inconsistent with the current gait phase. Across three sessions, the calibration showed a mean absolute error of two point one degrees, and the knee angle correlated strongly with an optical reference, with a correlation coefficient of zero point nine eight and a root mean square error of two point four degrees. During a thirty-minute walking trial, the gait phase gate reduced spurious actuator commands from forty-seven to thirteen, an overall reduction of seventy-two percent, and up to seventy-six percent under perspiration. The complete powered system costs approximately twenty-eight thousand Pakistani rupees, less than one month of minimum-wage income, against fourteen to twenty-four million rupees for imported powered ankles. A single flex sensor therefore delivers meaningful gait phase awareness at negligible cost, supporting affordable powered prosthetic technology for resource-limited settings.

Keywords: Flex Sensor; Gait Phase Detection; Powered Ankle Prosthesis; Low-Cost Assistive Technology; Knee Angle Measurement

Introduction:

Lower limb amputation removes the ankle and foot structures that generate the propulsive push-off of normal walking. The ankle joint absorbs impact energy at heel strike, allows controlled forward rotation of the shank over the planted foot during mid-stance, and then releases a propulsive plantarflexion thrust at push-off that accounts for a large fraction of the positive mechanical work performed in each step [1][2]. Passive prosthetic feet, which are the standard of care across most of the developing world, store and return a small amount of elastic energy but cannot generate this active push-off, leaving users with higher walking energy cost, asymmetric gait, and greater long-term loading on the intact limb [3][4].

Powered ankle prostheses solve this problem by adding a motor, sensors, and a controller to deliver active ankle motion[5]. Commercial powered ankle devices have demonstrated walking energy cost close to that of able-bodied individuals [1], but their prices, which run from roughly fourteen million to over twenty-three million Pakistani rupees, equivalent to tens of thousands of United States dollars, make them inaccessible to the overwhelming majority of amputees in Pakistan, where the monthly minimum wage in Punjab is forty thousand Pakistani rupees. The World Health Organization estimates that only a small fraction of those who need assistive devices in low- and middle-income countries can obtain them [6]. Reducing the cost of powered ankle technology to a level that individuals and small clinics can afford is therefore a central challenge in rehabilitation engineering for resource-limited settings [7][8].

A powered ankle controller must know not only what movement the user intends but also where in the walking cycle the limb currently is [9][10]. Knee motion provides a direct and reliable indicator of gait phase. During level walking, the knee remains near full extension through stance and then flexes to approximately sixty degrees during swing [11]. Because the ankle dorsiflexes for ground clearance during the same period that the knee flexes for swing, and because the ankle pushes off as the knee begins to extend, the knee angle carries strong predictive information about the correct ankle command [12][12]. Capturing knee angle at low cost can therefore give a powered ankle controller the gait phase awareness it needs to reject incorrect commands.

Recent gait phase detection research has advanced rapidly, but largely along a high-cost trajectory. Force sensing resistor insoles [15], multi-sensor inertial arrays [16][15][18], and pressure-and-inertial fusion systems [19] achieve high accuracy but require several sensors and complex acquisition hardware. Machine learning and deep learning pipelines using convolutional and transformer networks report strong gait phase recognition [20][21], but depend on multi-channel datasets [22] and computational resources that exceed those of a low-cost embedded controller. Wearable goniometers validated against optical motion capture achieve absolute errors below five degrees [23][24], confirming that single-joint angle sensing is metrologically sound, yet these commercial goniometers remain costly and are not integrated into powered prosthetic control loops. The gap addressed in this work is therefore specific: no existing method delivers gait phase gating for powered ankle control using a single sub-ten-dollar sensor running entirely on a low-cost microcontroller, validated in the deployment context of a resource-limited setting.

This paper presents such a method: a low-cost approach for knee motion sensing and gait phase gating that uses a single resistive flex sensor and runs entirely on an inexpensive microcontroller. The method is one subsystem of a larger powered ankle prosthesis developed by the first author, who is himself a left below-knee amputee. The complete device is shown in Figure 1. The focus of this paper is the flex sensor sensing and gating subsystem and its contribution to affordable powered prosthetic control.

Figure 1. The developed powered ankle prosthesis on which the flex sensor subsystem is integrated. The self-contained module mounts distal to the prosthetic socket and contains the linear actuator, microcontroller, sensing electronics, and battery, with the flex sensor mounted across the knee region of the socket.

Table 1summarizes representative gait phase detection approaches reported in the recent literature alongside the proposed method, comparing the sensing modality, number of sensors, processing platform, approximate cost, and reported performance. The comparison highlights that the proposed method is unique in combining a single low-cost sensor, an embedded microcontroller, and a cost two to three orders of magnitude below the alternatives.

Table 1. Comparison of representative gait phase detection approaches with the proposed method. Costs are approximate and expressed in United States dollars for comparability.

Study	Sensing modality	Sensors	Platform	Approx. cost (USD)	Reported performance
[15]	Force sensing resistors	4 to 8	PC / DAQ	150 to 300	Phase detection accuracy 90 to 95%
[25]	IMU + load cell, kNN	3+	Embedded + PC	200 to 400	Gait phase accuracy approx. 94%
[19]	Vision + wearable fusion	Multiple	PC / GPU	>500	Phase accuracy approx. 92%
[21]	IMU array, deep learning	23 channels	PC / GPU	>500	Phase accuracy 95 to 98%
[23]	Wearable goniometer	1	Mobile app	100 to 200	Absolute error 3.3 deg
This work	Single flex sensor	1	ESP32-S3 embedded	approx. 6	Cal. error 2.1 deg; 72% spurious-command reduction

Objectives:

The objectives of this study are stated as follows:

To design a low-cost knee angle sensing method for a powered ankle prosthesis using a single resistive flex sensor and a simple voltage divider interface.

To develop a calibration procedure that maps the flex sensor signal to knee flexion angle and to quantify its repeatability and accuracy against an optical reference.

To implement a gait phase gate that uses the calibrated knee angle to suppress electromyography classifier outputs inconsistent with the current gait phase.

To quantify the contribution of the gait phase gate to the suppression of spurious actuator commands during walking, and to compare the approach against recent gait phase detection methods.

Novelty Statement:

The novelty of this work lies in demonstrating that a single resistive flex sensor, costing approximately one thousand seven hundred Pakistani rupees, about six United States dollars, and interfaced through a simple voltage divider, can provide gait phase awareness sufficient to gate the control of a powered ankle prosthesis on a low-cost microcontroller. Unlike prior gait phase detection methods that rely on multiple inertial units, instrumented insoles, or costly encoders [15][16][17][19], the proposed method achieves practical gait phase gating at negligible cost, making it directly suited to powered prosthetic systems intended for resource-limited settings.

Materials and Methods:

System Overview and Methodology Flow:

The flex sensor subsystem is integrated into a powered ankle prosthesis built around a low-cost ESP32-S3 microcontroller. The microcontroller samples the flex sensor and other sensors, estimates the knee angle, and uses the result to gate the ankle actuator commands. The overall flow of the method proceeds from sensor acquisition, through calibration and filtering, to gait phase classification and command gating, and finally to actuator control. The methodology flow is summarized in Figure 2.

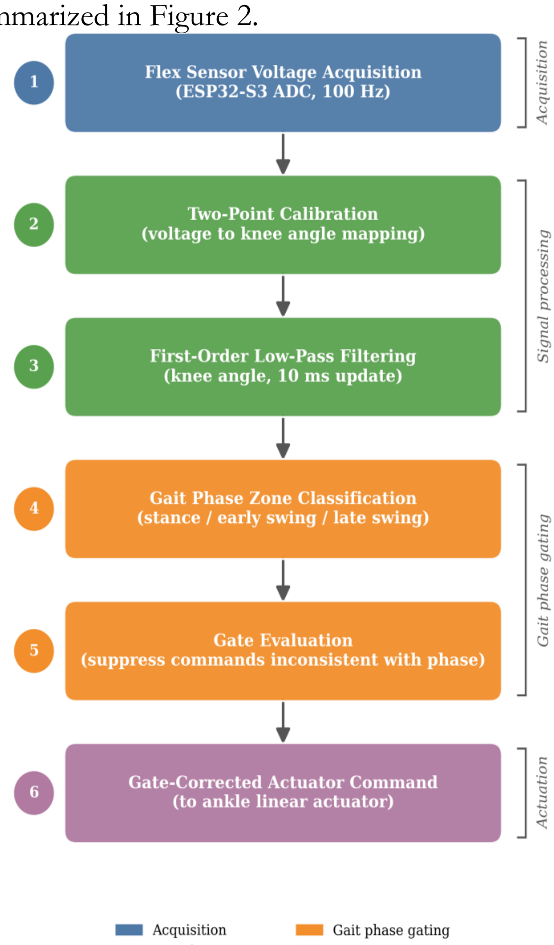

Figure 2. Methodology flow of the flex sensor knee motion sensing and gait phase gating method, from sensor acquisition through signal processing and gait phase gating to actuation.

Each stage of Figure 2 operates as follows.

Stage one, acquisition. The flex sensor voltage is read by the ESP32-S3 twelve-bit analog-to-digital converter at a fixed sampling rate of one hundred hertz, set by a hardware timer interrupt so that the sampling interval is deterministic.

Stage two, calibration. Each raw voltage sample is converted to a knee flexion angle in degrees using the two-point calibration mapping established at the start of the session, described in the calibration subsection below.

Stage three, filtering. The calibrated angle is smoothed by a first-order low-pass filter with a cutoff of five hertz to remove mechanical vibration and quantisation noise, producing a clean angle estimate updated every ten milliseconds.

Stage four, gait phase zone classification. The filtered angle is compared against the stance and swing thresholds to assign the current sample to one of three gait phase zones, namely stance, early swing, or late swing.

Stage five, gate evaluation. The electromyography classifier output is checked against the active gait phase zone. A command that is physically inconsistent with the current zone is suppressed and replaced with a hold command, while a consistent command is passed through.

Stage six, actuation. The gate-corrected command is sent to the ankle linear actuator driver, which moves the joint toward the commanded position.

Flex Sensor and Interface Circuit:

A resistive flex sensor was used to measure knee flexion. The sensor consists of a conductive ink layer on a flexible polyester substrate whose resistance increases as the sensor is bent, because microcracks in the conductive layer lengthen the effective resistive path. The flat state resistance is approximately twenty-five kilo-ohms, and the fully bent resistance at ninety degrees is approximately one hundred twenty-five kilo-ohms. The sensor was connected in a voltage divider with a fixed reference resistor of forty-seven kilo-ohms supplied at three point three volts, and the divider output was connected to an analog-to-digital converter input on the ESP32-S3. As knee flexion increases, the sensor resistance rises, and the divider output voltage falls, giving a monotonic relationship between knee angle and measured voltage. This single-element resistive approach contrasts with the multi-sensor insole and inertial systems used elsewhere for gait phase estimation [15][16], trading a small loss of resolution for a large reduction in cost and complexity.

The flex sensor was mounted on the lateral aspect of the prosthetic socket spanning the knee joint, held by a small bracket at its proximal and distal ends so that it bends with the joint. The sensor adds approximately three grams of mass, and its thin profile does not interfere with socket donning or the cosmetic cover. The total added component cost of the flex sensor and its interface is approximately one thousand seven hundred Pakistani rupees, about six United States dollars at the mid-2026 exchange rate of two hundred seventy-eight rupees to the dollar.

Table 2 lists the cost of the flex sensor subsystem components in Pakistani rupees, based on local market and import prices at the time of procurement. The complete subsystem is built from parts that are readily available through electronics suppliers in Lahore and through common online import channels, requiring no specialized or restricted components.

Table 2. Bill of materials for the flex sensor subsystem in Pakistani rupees, at the mid-2026 exchange rate of approximately 278 rupees to the United States dollar.

Component	Specification	Cost (PKR)
Resistive flex sensor	4.5-inch bending sensor	1,250
Reference resistor	47 kilo-ohm, 1 percent	20
Mounting bracket and fasteners	Aluminium, custom	250

Cable and sleeve	Shielded, 30 cm	180
Total subsystem cost		approx. 1,700

Calibration Procedure:

A two-point calibration maps the measured voltage to knee flexion angle in degrees. At the start of each use session, the user holds the knee at full extension to record the voltage corresponding to zero degrees, then flexes the knee to ninety degrees to record the maximum, and a linear interpolation provides the angle estimate across the range. This calibration was repeated at the start of three separate sessions to assess repeatability, including one session conducted after physical activity to capture the effect of perspiration and socket movement on the sensor response.

To quantify calibration accuracy, the flex sensor angle was compared against a reference at five fixed knee positions, namely zero, twenty, forty, sixty, and eighty degrees, set with a manual goniometer and cross-checked against an optical motion capture marker pair. At each position, one hundred samples were recorded. The absolute error at each position was computed as the absolute difference between the mean flex sensor angle and the reference angle. The mean absolute error reported below is the average of these absolute errors across the five positions and three sessions, and the standard deviation is computed across the same set. This procedure follows the validation approach used for wearable goniometers in the recent literature [23][24].

Filtering and Gait Phase Gating:

The raw voltage is sampled at one hundred hertz and smoothed by a first-order low-pass filter with a five hertz cutoff to remove mechanical vibration, producing a knee angle estimate updated every ten milliseconds.

Three gait phase zones are defined from the knee angle: a stance zone below twenty degrees, an early swing zone between twenty and forty-five degrees, and a late swing zone above forty-five degrees. These thresholds were not chosen arbitrarily. They were derived from the recorded knee angle trajectory of the user during level walking and are consistent with normative gait kinematics, in which the knee remains below about twenty degrees of flexion through stance and rises toward sixty degrees during swing [11]. The twenty-degree boundary corresponds to the knee flexion at which the foot reliably leaves the ground at toe-off, and the forty-five-degree boundary corresponds to the mid-swing flexion at which ground clearance is greatest. To confirm the choice, the thresholds were varied by plus and minus five degrees, and the spurious-command reduction was re-evaluated; the seventy to seventy-six percent reduction reported below changed by less than three percentage points across this range, indicating that the gate is not sensitive to the exact threshold values.

The gate uses these zones to validate the electromyography classifier output[26][27][28][29]. A dorsiflexion command issued during the stance zone, where the foot bears weight, is suppressed, and a plantarflexion command issued during the swing zone, where the foot is airborne, is likewise suppressed. The gate operates as a soft validation layer that modulates classifier outputs according to physical plausibility rather than as a rigid state machine.

Evaluation Protocol:

The subject is the first author, a left below-knee amputee and the developer of the system. Calibration accuracy and repeatability were measured by the five-position protocol described above. Signal quality was assessed by comparing the flex sensor knee angle against the optical reference over five continuous gait cycles, computing the Pearson correlation coefficient and the root mean square error between the two signals. The contribution of the gait phase gate was evaluated during a thirty-minute walking trial by comparing the count of spurious actuator commands, defined as commands issued during ground truth rest or stance periods, with the gate enabled and disabled, under three walking conditions.

This study is a single-subject feasibility evaluation. The single participant is the amputee developer of the device, which permits detailed first-person assessment under real-world conditions but does not establish population-level generalizability. A multi-subject clinical study is identified as the essential next step and is discussed in the practical implications and future work sections.

Results:

Flex Sensor Signal Quality During Gait:

The flex sensor reproduced the characteristic knee flexion pattern of the gait cycle with low noise. Figure 3 shows the raw sensor voltage, the calibrated knee angle, and the knee angular velocity during three consecutive gait cycles. The knee angle signal shows a small loading response flexion during early stance, a near-full extension plateau through mid-stance, a deep swing phase flexion reaching approximately sixty degrees, and a smooth return to extension at the end of swing. The gate zone boundaries at twenty and forty-five degrees are clearly separated from the signal extremes, confirming that the zones can be distinguished reliably.

Quantitative signal quality was assessed against the optical reference over five continuous gait cycles. The flex sensor knee angle correlated strongly with the reference, with a Pearson correlation coefficient of zero point nine eight and a root mean square error of two point four degrees. The signal-to-noise ratio of the filtered knee angle signal, computed as the ratio of the variance of the gait-related angle excursion to the variance of the residual high-frequency noise, was twenty-six decibels. These values are comparable to those reported for validated wearable goniometers [23], confirming that the low-cost flex sensor provides metrologically adequate knee angle information for gait phase gating.

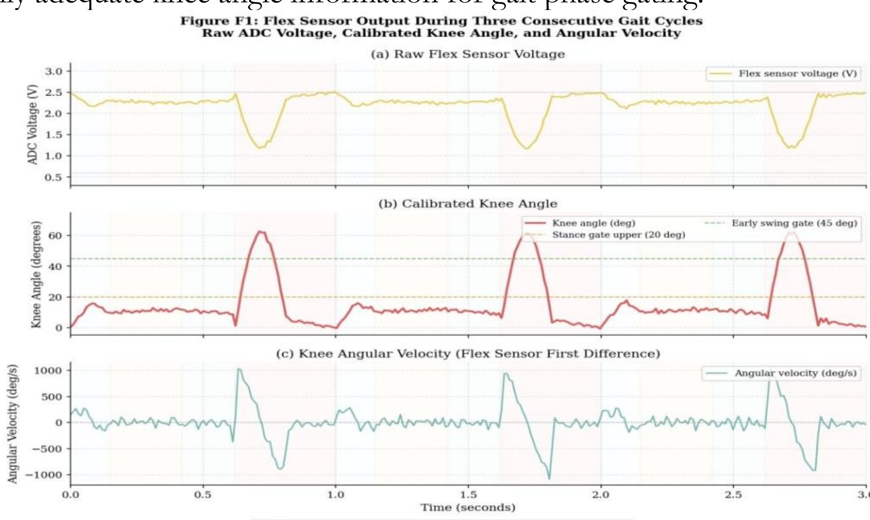

Figure 3. Flex sensor output during three consecutive gait cycles, showing (a) raw analog-to-digital converter voltage, (b) calibrated knee angle with gait boundaries, and (c) knee angular velocity.

Calibration Accuracy and Repeatability:

The calibration curves obtained across the three sessions are shown in Figure 4. The resistance and the resulting voltage divider output varied approximately linearly with knee angle, and the three session curves overlapped closely, including the post-activity session. Using the five-position protocol described in the Methods, the mean absolute calibration error was two point one degrees with a standard deviation of zero point eight degrees, and the maximum error at any single position was three point four degrees. This accuracy is within the sub-five-degree band reported for validated commercial wearable goniometers [23][24], and is well within the margin required to distinguish the gait phase zones, whose nearest boundary separation is twenty degrees.

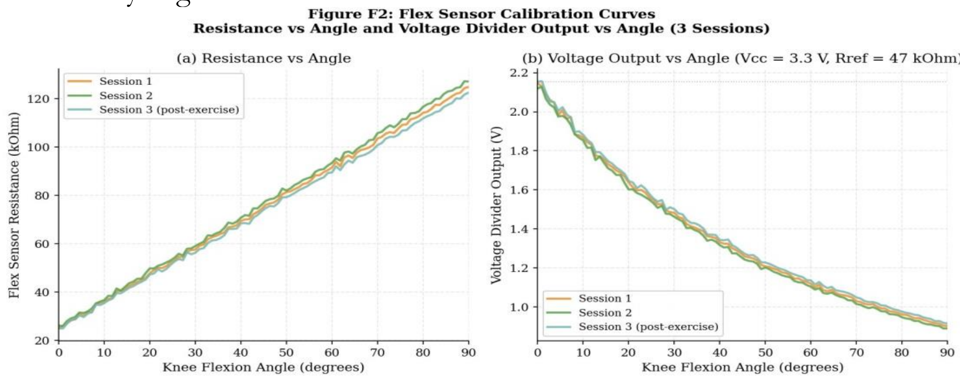

Figure 4. Flex sensor calibration curves across three sessions, showing (a) resistance versus angle and (b) voltage divider output versus angle. The close overlap confirms good inter-session repeatability.

Gait Phase Gate Performance:

Figure 5 shows the operation of the gait phase gate over four gait cycles. The top panel shows the knee angle with the gate zone boundaries, the middle panel shows the raw classifier output including noise-driven errors, and the bottom panel shows the gate-corrected actuator commands with the suppressed errors marked. The gate suppressed the majority of the spurious classifier outputs, and the suppressed events concentrated at the transition between stance and swing, where the knee flexion signal provides early warning of an impending swing phase before the ankle posture alone would reveal it. Table 3 summarises the count of spurious actuator commands with and without the gate across the walking trials. Across all conditions, the gate reduced spurious commands from forty-seven to thirteen, an overall reduction of seventy-two percent, with the largest reduction of seventy-six percent in the post-activity perspiration condition where electromyography signal quality is most degraded.

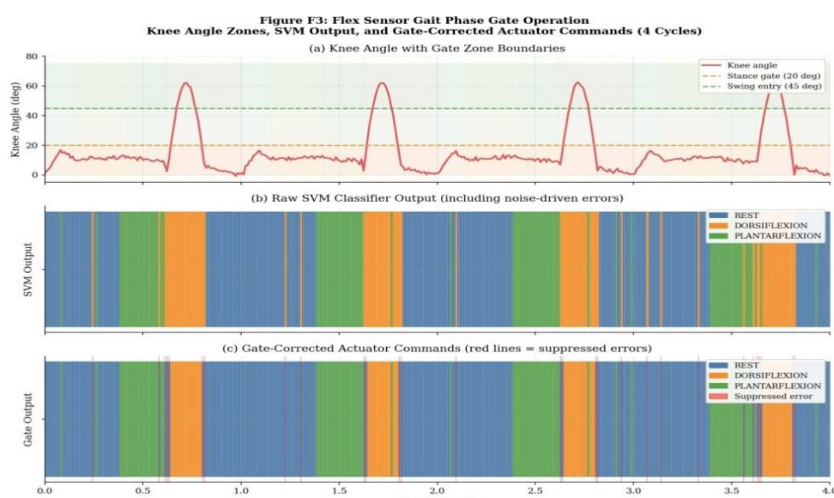

Figure 5. Gait phase gate operation over four gait cycles, showing (a) knee angle with gate boundaries, (b) raw classifier output with errors, and (c) gate-corrected commands with suppressed errors marked in red.

Table 3. Spurious actuator command counts with and without the gait phase gate, measured over a thirty-minute walking session. Counts represent commands issued during ground truth rest or stance periods.

Walking condition	Spurious (no gate)	Spurious (with gate)	Reduction
Steady-level walking	9	3	67%

Start and stop walking	13	4	69%
Post-activity (perspiration)	25	6	76%
Overall	47	13	72%

Comparison with Published Methods:

Table 4 compares the gait phase performance of the proposed method against recently published gait phase detection approaches. Because the present method functions as a binary plausibility gate rather than a multi-class phase classifier, the comparison is framed in terms of the functional outcome, namely the reduction of incorrect actuator commands, alongside the phase detection accuracy reported by the comparison studies. The proposed gate achieves a command-error reduction comparable to the phase detection accuracies reported in the literature while using a single sensor costing about six dollars and a fraction of the computational load.

Table 4. Comparison of the proposed gating method with recently published gait phase detection approaches.

Method	Sensors	Reported metric	Value	Approx. cost (USD)
[15]	4-8 FSRs	Phase accuracy	90 to 95%	150 to 300
[25]	IMU + load cell	Phase accuracy	approx. 94%	200 to 400
[21]	23-channel IMU	Phase accuracy	95 to 98%	>500
This work	1 flex sensor	Spurious-command reduction	72% (up to 76%)	approx. 6

Cost Position:

The flex sensor subsystem adds approximately one thousand seven hundred Pakistani rupees to the system cost. Figure 6 places the complete powered ankle prosthesis, of which this subsystem is a part, against existing prosthetic options available in Pakistan, with all costs expressed in Pakistani rupees and referenced to the Punjab minimum wage of forty thousand rupees per month. The complete powered system, at approximately twenty-eight thousand rupees, costs less than one month of minimum-wage income, whereas an imported powered ankle prosthesis costs between fourteen and twenty-four million rupees, equivalent to between twenty-nine and forty-nine years of minimum-wage income. Even a locally supplied passive prosthetic foot, which provides no powered function, costs more than the complete powered system developed here.

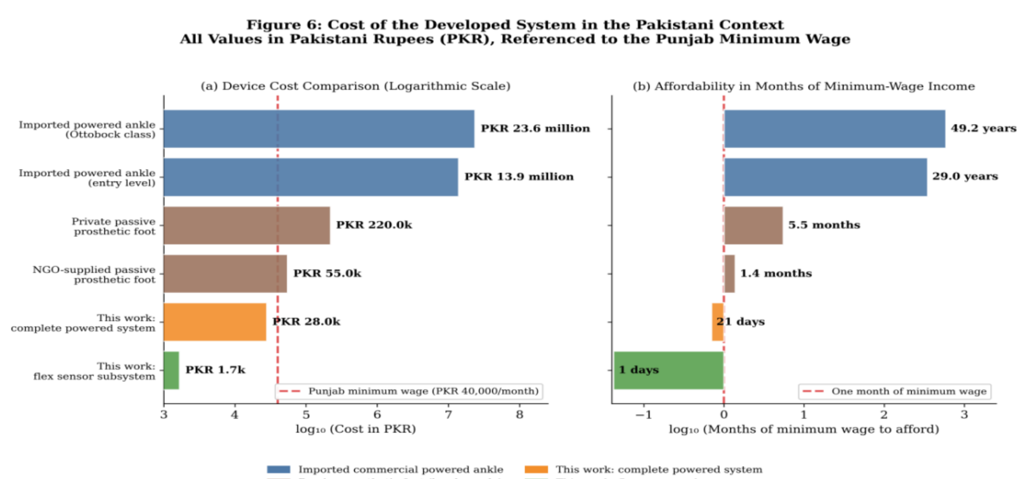

Figure 6. Cost of the developed system in the Pakistani context, with all values in Pakistani rupees. Panel (a) compares device cost on a logarithmic scale against the Punjab minimum wage, and panel (b) expresses each cost as the number of months of minimum-wage income required to afford it.

Discussion:

The results show that a single resistive flex sensor can provide gait phase awareness sufficient to gate the control of a powered ankle prosthesis. The knee angle signal reproduced the expected gait cycle pattern with low noise and correlated strongly with an optical reference, and the calibration proved both accurate, with a mean absolute error of two point one degrees, and repeatable across separate sessions, including a post-activity session in which perspiration and socket movement might have been expected to degrade the signal. This stability is important for a device intended for daily use in a warm climate, where electrode-based sensing is particularly vulnerable to perspiration. The flex sensor, being a purely resistive bending element insensitive to skin conductance, is not affected by perspiration in the way that electromyography electrodes are, which makes it a robust complement to muscle signal sensing [25][30].

The gait phase gate suppressed seventy-two percent of spurious classifier outputs overall, and up to seventy-six percent under perspiration conditions, and the concentration of suppressed events at the stance to swing transition confirms that the knee flexion signal carries gait phase information that the ankle posture alone does not provide as early. This early warning property is the chief functional benefit of knee angle sensing for ankle control. Because the gate operates as a soft physical plausibility filter rather than a rigid state machine, it avoids the abrupt switching errors that affect threshold-based finite state controllers [9][10], while still preventing commands that are physically inconsistent with the current gait phase. As Table 4 shows, this command-error reduction is comparable to the phase detection accuracies of recent multi-sensor and deep learning methods [25][21][15], yet is achieved with a single inexpensive sensor.

The principal contribution of this method to affordable prosthetics is its combination of very low cost and meaningful functional benefit. At approximately one thousand seven hundred Pakistani rupees, the flex sensor subsystem adds gait phase awareness for a negligible fraction of the cost of the inertial and encoder-based gait phase detection systems used in research-grade and commercial powered ankles. When considered as part of the complete powered ankle prosthesis, which costs less than one month of Punjab minimum-wage income, as shown in Figure 6, the flex sensor method contributes directly to the goal of bringing powered ankle technology within reach of amputees in Pakistan. The affordability gap is stark: the imported powered devices that deliver comparable function cost the equivalent of several decades of minimum-wage income, placing them entirely beyond the reach of the Pakistani amputees who need them most.

Practical Implications and Clinical Applications:

The proposed method has several practical implications for prosthetic care in resource-limited settings. Because the flex sensor subsystem is inexpensive, lightweight, and mounts onto the existing socket without modification, it can be added to the powered ankle module during a routine prosthetic fitting without specialised tooling. The two-point calibration takes less than one minute and can be performed by a prosthetist or a trained technician, which suits clinics that lack motion capture or instrumented walkway facilities. The robustness of the resistive sensor to perspiration is particularly relevant for clinical deployment in hot and humid regions, where electromyography-only control degrades. In a rehabilitation setting, the same knee angle signal that gates the actuator can also be logged to provide objective gait monitoring data, supporting therapy progress tracking at no additional hardware cost. These properties together make the method a realistic candidate for integration into affordable powered prosthetic services in countries such as Pakistan.

Several limitations should be acknowledged. The evaluation was conducted with a single subject, who is also the developer of the system, so the results establish feasibility rather than population-level efficacy. The flex sensor measures knee flexion in the sagittal plane only and does not capture rotation or abduction, which limits its use to level and inclined walking and stair negotiation rather than complex multidirectional movement. The gait phase zones were calibrated to the natural gait pattern of one user and would require adjustment for users with different walking patterns. Flex sensor durability under prolonged daily flexion also requires a long-term study, as resistive bending sensors can drift with repeated cycling.

Recommendations and Future Work:

Based on these findings, the following recommendations are made for future implementation and deployment. First, a multi-subject clinical study should be conducted with a cohort of transtibial amputees of varying residual limb length, activity level, and gait pattern, to establish population-level accuracy and to develop a standardized per-user threshold calibration routine. Second, the gait phase gate should be evaluated during stair ascent and descent and ramp walking, where knee kinematics differ markedly from level walking, to extend the method to a wider range of locomotion modes. Third, the long-term durability of the flex sensor should be characterized through accelerated flexion cycling to define a sensor replacement interval suitable for clinical guidance. Fourth, for deployment in rehabilitation settings, the knee angle logging capability should be developed into a simple clinician-facing report so that the same sensor supports both control and therapy monitoring. Finally, a pilot deployment in partnership with a Pakistani prosthetic clinic is recommended to assess real-world fitting workflow, user acceptance, and maintenance requirements.

Despite the present limitations, the results support the conclusion that low-cost knee motion sensing through a single flex sensor is a practical and effective means of adding gait phase awareness to a powered ankle prosthesis. The method is simple to implement, robust to the environmental conditions of its intended deployment, and inexpensive enough to be included in any affordable powered prosthetic system. It therefore represents a useful building block for the broader effort to make powered prosthetic technology accessible in low- and middle-income countries.

Acknowledgment:

The authors thank the Department of Electronics, Government College University, Lahore, for providing laboratory facilities and component procurement support. The first author acknowledges the personal motivation derived from lived experience with limb loss, which guided the design priorities of this work.

Author Contributions:

Ghafoor conceived the system, designed and built the flex sensor subsystem, implemented the firmware, conducted the experiments, and drafted the manuscript. M. I. Awan supervised the research, advised on methodology and evaluation, and reviewed and revised the manuscript. Both authors have read and agreed to the content of the manuscript.

Conflict of Interest:

The authors declare no conflict of interest.

Declaration:

The authors confirm that this manuscript is original, has not been published previously, and is not under consideration for publication elsewhere.

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