Untethered Wearable Loop Sensor System for Monitoring Human Joint Movement

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***Abstract*—We report a new class of wearable loop sensors for monitoring human kinematics (particularly, joint flexion angles) while overcoming limitations in the state-of-the-art. Previous studies have demonstrated the feasibility of these loop sensors using tethered connections to a network analyzer. In this work, we take a major step forward to demonstrate untethered operation for the sensor. To this end, transmitter and receiver boards are designed and integrated into the loops. The transmitter board sends a Radio-Frequency (RF) power of 5.68 dBm at 34 MHz upon a 50 Ω load, while the receiver board detects the power level and transmits the data to a nearby personal computer (PC) via Bluetooth. Flexion tests are conducted upon a tissue-emulating phantom to validate the setup. To quantify performance, we calculate the root mean square error (RMSE) between the estimated angle from our sensor and the gold-standard angle from a marker-based motion capture camera system, as well as Pearson's correlation coefficient (ρ). The proposed sensor shows outstanding performance with an average RMSE of 0.670° and an average ρ of 0.99966. Overall, our sensor outperforms state-of-the-art wearable kinematic technologies by being highly accurate, seamless, lightweight, unobtrusive to natural motion, and reliable over time.**

I. Introduction

Accurate monitoring of human joint kinematics offers significant benefits across diverse fields such as healthcare (including prevention [1], rehabilitation [2], and training [3][11]), sports analytics, and virtual reality. For example, individuals suffering from anterior cruciate ligament (ACL) injury [1] and mild traumatic brain injury (mBTI) [4] are at a high risk of subsequent injury due to neuromuscular and musculoskeletal disorders. The ability to monitor kinematics in real-world environments (i.e., outside of the clinic or lab) and at clinical-grade accuracy would be game-changing for these applications and beyond.

Optoelectronic motion capture (MoCap) systems are considered as today's "gold standard" for human kinematic analysis [5]. In this case, a set of cameras observes the location of limbs and the collected data are post-processed to extract joint angles. Although MoCap systems provide high accuracy, their inability to monitor real-life kinematics has led to a growing interest in wearable sensors for tracking daily activities. Such wearable kinematics sensors are known to operate based on either direct or indirect operating principles.

The direct sensing method, which includes fiber-optic sensors [6][7] and bending sensors (resistive [8] and capacitive [9]), involves placing the sensors directly on the joint to capture the angle using flexible materials. Fiber-optic sensors bend as the joint flexes, but are known to hamper natural movement. Bending sensors can bend and stretch effectively with joint movements. However, resistive bending sensors suffer from hysteresis effects due to stretching and bending deformations, while capacitive bending sensors experience reduced accuracy because of the capacitive effect between the sensors and the skin. Additionally, direct attachment to the human joint can result in discomfort for the users while performing daily life activities.

By contrast, indirect sensing methods rely on calculating the relative position of sensors placed on the limb rather than on the joint. Compared to the direct method, this approach improves comfort and does not restrict movement. Inertial measurement units (IMUs) [10][11] and time-of-flight sensors are the main examples. IMUs, which typically integrate an accelerometer, gyroscope and magnetometer, derive angles by integrating raw acceleration data. However, they suffer from integration drift, as errors accumulate over time. To address this issue, sophisticated algorithms (such as Kalman filter) must be applied to continuously calibrate the sensors. Time-of--flight sensors, which include electromagnetic-based [12] and acoustic-based [13] types, calculate angles unitizing the law of cosines by knowing the distance between the transmitting (Tx) and receiving (Rx) sensors as well as the distance between the joint and the Tx/Rx sensors. However, they are prone to interference, resulting in low reliability for daily usage.

Inspired by the indirect sensing method, we recently proposed electromagnetic-based wearable loop sensors that include Tx and Rx loops placed symmetrically across the joint to capture sagittal plane kinematics, e.g., the loops would be placed on the shank and thigh when capturing sagittal knee flexion [14]-[16]. The two loops operate in the deep induction region base on Faraday's Law, with a unique transmission coefficient ($|S_{21}|$) value for each knee flexion angle. In our previous work, the sensors were connected to a network analyzer to record $|S_{21}|$, which was then post-processed based on a calibration map to derive the knee flexion angle. Our proposed sensors are proven to be reliable over time, lightweight, comfortable for users, and robust to noise inherent to natural environments. However, connection to a network analyzer compromises portability, thus confining usage of our previously reported prototypes to laboratory environments.

To overcome this limitation, we herewith take a major step forward to develop an untethered system for monitoring – without loss of generality – sagittal plane kinematics. The system includes our previously reported wearable loop sensors, along with novel Tx and Rx boards for signal transmission, reception, and Bluetooth communication. Experiments are conducted on a tissue-emulating phantom using the untethered system and a gold-standard camera system for comparison. For the first time, this work confirms the feasibility of monitoring flexion/extension angles with wearable loop sensors in an untethered setup.

II. Materials and Methods

A. Sensor Operating Principle

Per Figure 1(a), one Tx loop and one Rx loop of 8 cm in radius are placed symmetrically across the joint (in this case, the knee). The two loops operate in the deep induction region at 34 MHz and are designed to be resonant for maximum power transfer. According to Faraday's Law of induction, the magnetic flux on the Rx loop changes with joint flexion/extension, resulting in varying received power levels for different flexion angles. Figure 1(b) plots the relationship between the joint flexion angle (θ_f per definition in Figure 1(a)) and $|S_{21}|$ (with ports 1 and 2 corresponding to the Tx and Rx loops, respectively). As seen, for every flexion angle, there is a uniquely corresponding value of $|S_{21}|$. That is, by collecting real-time $|S_{21}|$ data, we can determine the flexion angle based upon a predefined relationship that can be retrieved at the calibration stage.

B. Circuit Operating Principle

We propose a wireless and untethered sensor system that includes the loop sensors of Figure 1(a), a Tx board to send the 34 MHz signal, and a Rx board to collect data and send them to a personal computer (PC) through Bluetooth, as shown in Figure 1(c). The Tx board consists of a CMOS oscillator (SiT80008BC, SiTime) and a fifth-order Chebyshev low-pass filter (LPF) with a 40 MHz cutoff frequency. It generates radio frequency (RF) power of $P_t = 5.68$ dBm at 34 MHz, terminated with a 50 Ω load. The Rx board consists of a bandpass filter with a cutoff frequency from 30 MHz to 40 MHz (SXBP-35N+, Mini-Circuits), a logarithmic amplifier (ADL5513, Analog Devices), and a 12-bit analog-to-digital converter with a Bluetooth module (CYBLE-012011-00, Infineon). It can detect received power levels (P_r) from -70 dBm to 10 dBm with an 80 dB dynamic range; convert the power to voltage; and wirelessly transmit the voltage data to a PC via Bluetooth. As would be expected, the Tx loop is connected to the Tx board, and the Rx loop is connected to Rx board. The RF power generated by the Tx board is coupled from the Tx loop to the Rx loop, and then received and processed by the Rx board.

According to the theory behind loop sensors outlined in Section II.A, the relationship between $|S_{21}|$ and flexion angle is unique. This implies that the relationship between the received power and flexion angle is also unique. Given that the logarithmic amplifier establishes a one-to-one correspondence between the received power and output voltage, we can determine the flexion angle from the collected voltage data on the Rx board.

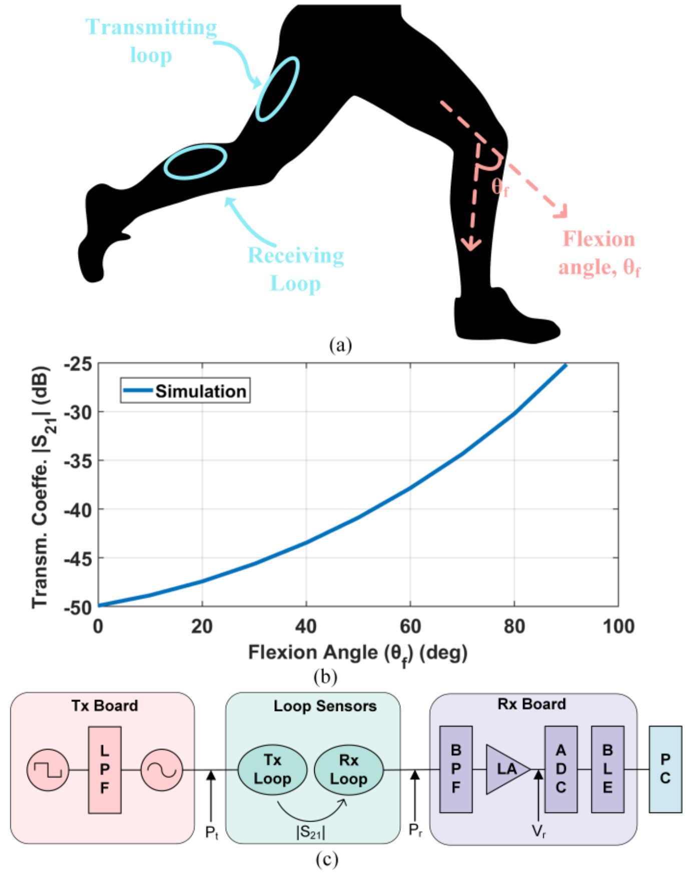


Figure 1. (a) Schematic of the wearable loop sensors symmetrically placed across the knee joint. (b) Relationship between the joint flexion angle (θ_f per definition in Figure 1(a)) and $|S_{21}|$. (c) Block diagram of the proposed untethered sensor system, including the Tx board, loop sensors and Rx board (LPF: low-pass filter, BPF: band-pass filter, LA: logarithmic amplifier, ADC: analog-to-digital converter, BLE: Bluetooth low energy, PC: personal computer).

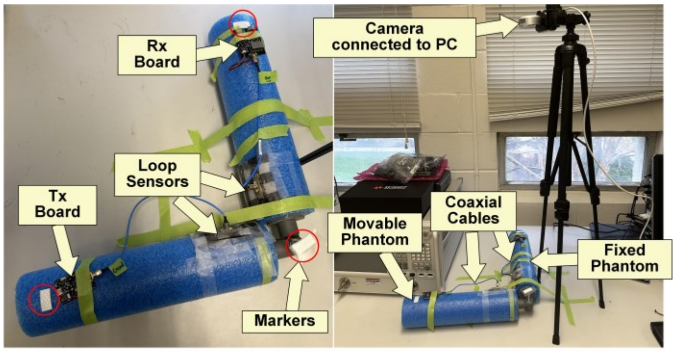
C. Experimental Setup

Figure 2(a) shows the top view of the untethered wearable loop sensor system on a tissue-emulating Styrofoam model. We note that since tissues are non-magnetic and sensors are operating in the deep induction region, there is no need to mimic the actual biological tissues. The Styrofoam limb, which has a diameter of 8 cm, is connected to a 3D-printed joint to mimic joint flexion and extension. The Tx and Rx loops are connected to the Tx and Rx boards, respectively, using coaxial cables. The complete experimental setup is shown in Figure 2(b). A marked-based MoCap system (Intel RealSense515 LiDAR camera) is utilized to retrieve the gold-standard angles for comparison. Data from both the wearable sensor system and the camera are collected on a PC.

D. Data Collection Process

We first examined the linearization performance of the logarithmic amplifier on the Rx board. The received power (P_r)

from the loop sensor and the received voltage from the logarithmic amplifier were recorded using a spectrum analyzer (Keysight N9020B) and PC for joint flexion angles that varied from 0 to 90 degrees in 10-degree increments.



(a) (b)

Figure 2. (a) Top view of the untethered wearable loop sensor system. (b) Complete experimental setup showing the wearable sensor and MoCap camera.

Calibration was conducted with continuous slow flexion and extension movements from 0 to approximately 90 degrees, followed by several fast flexion and extension movements within a period of 60 seconds. The purpose of the slow flexion and extension is to establish the relationship between the received voltage and flexion angle as needed for data analysis while avoiding potential noise caused by motional electromotive force (EMF) as discussed in [16]. Fast flexion and extension movements are intended to align the time stamp for the sensor and camera data. This calibration process was repeated three times to ensure repeatability.

After calibration, we performed manual flexion and extension movements at random speed for 80 seconds. We repeated this process five times and captured data using both our sensor and camera. Time stamps for both datasets were aligned using peak-to-peak alignment. We analyzed the data from 10 seconds to 60 seconds in which both datasets were complete and aligned in time.

III. Experimental Results

As shown in Figure 3(a), we tested the linearity of the logarithmic amplifier using ten discrete data points. The blue solid line represents the expected relationship between the input (received power) and output (received voltage), derived by applying linear regression based on the data points at both ends. The red dots represent the data points at different flexion angles, as mentioned in Section II.D. As seen, all red dots are located on the blue solid line, illustrating the linearity of the selected logarithmic amplifier and proving the feasibility of our proposed untethered system.

Figure 3(b) shows the calibration curve that connects the received voltage from our sensor with the flexion angle from the MoCap camera. Results for the three experiments overlap with each other, confirming the accuracy of our calibration results. Using this calibration curve, we can transform the voltage data collected during the experiment into estimated angle data.

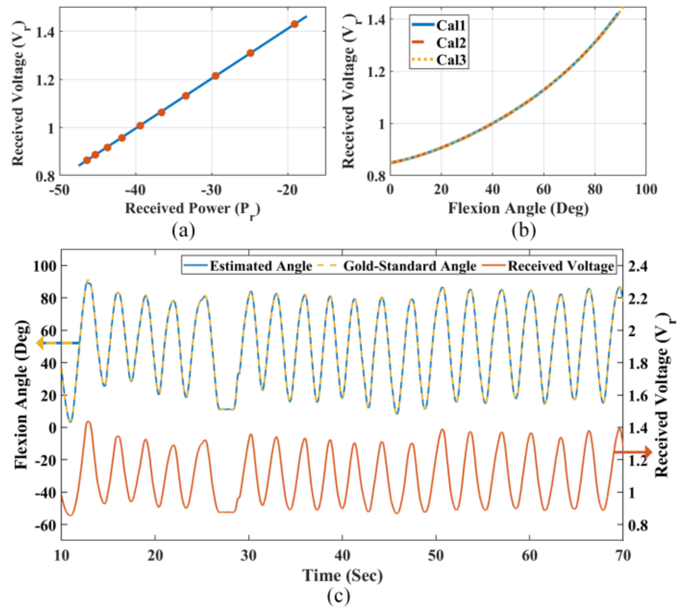


Figure 3 (a) Linearization test results for the logarithmic amplifier. (b) Calibration result for the untethered wearable loop system (c) Representative experiment result for the untethered wearable loop system.

Figure 3(c) shows an example set of experimental data. The red solid line, yellow dashed line, and blue solid line represent the received voltage from our sensor, the gold-standard angle from the MoCap camera, and the estimated angle derived from the received voltage based on the calibration curve, respectively. The estimated angle and the gold-standard angle nearly overlap, proving the high accuracy of our proposed sensor.

To quantify the sensor's performance, we calculated the root mean squared error (RMSE) and Pearson's correction coefficient (ρ) between the estimated angle (from our sensor) and the gold-standard angle (from the camera) for each trial. Results are shown in Table I. Compared with state-of-the-art wearable sensors that exhibit an RMSE no lower than 3°, our sensors achieve extremely low average RMSE of 0.670° ± 0.366°.

TABLE I
Experimental Results for Flexion Angle Obtained via the Untethered Loop Sensor System

Trial Number	RMSE	ρ
1	0.373°	0.9999
2	1.025°	0.9994
3	0.618°	0.9997
4	0.495°	0.9998
5	0.842°	0.9995
Average	**0.670° ± 0.366°**	**0.99966±0.00026**

IV. Conclusion

In this paper, we reported an untethered loop sensor system for wearable monitoring of human joint kinematics. Feasibility of the wearable loops has been demonstrated in previous works using tethered connections to a network analyzer. Here, we enhanced the sensor's portability by integrating transmitting and receiving boards with Bluetooth connection to a remote computer. Accuracy was validated on a phantom model, achieving an RMSE of 0.670° ± 0.366° that considerably outperforms the state-of-the-art. Additionally, the sensor is easy to integrate into clothing to collect daily activity data without hampering natural movement. We envision a future where clinicians can use this sensor to collect kinematics in real-world environments.

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