A Novel Flexible Wearable Sensor for Estimating Joint-Angles

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***Abstract*— To circumvent current limitations of wearable sensors that can be used to assess and monitor joint movements, we developed an accurate, low-cost, flexible wearable sensor comprising a retractable reel, a string, and a potentiometer. This sensor is intended to estimate joint angles in correlation with the amount of skin stretch measured by the change in the length of the string. In this study, we validated the accuracy of the sensor against an optoelectronic system in estimating knee joint angles using a dataset obtained from 9 healthy individuals while they walk and run on a treadmill. By our simple calibration procedure, we could convert the voltage output of the potentiometer to the amount of skin stretch as subjects flex or extend their knee. Then, we incorporated a simple polynomial fitting model to estimate the joint angle. Using a leave-one-subject-out cross validation, we achieved an average root mean square error of 4.51 degrees. This work demonstrates the accuracy of the proposed system in estimating knee joint angles and provides the basis to develop more complex systems to assess and monitor joints having more degrees of freedom. We believe that our novel low-cost wearable sensing technology has great potential to enable joint kinematic monitoring in ambulatory settings.**

I. INTRODUCTION

Modeling and monitoring of human movement in ambulatory settings has recently received significant attention in many disciplines including clinical and health sciences [1], wellness [2], smart living [3], robotics [4], human computer interaction [5], and entertainment [6]. For example, in clinical and health sciences, understanding kinematic parameters such as joint range of motion or various gait parameters has been of great interest for tracking disease progress or evaluating the efficacy of clinical interventions (e.g. rehabilitation) in many neuromuscular and musculoskeletal disorders. Clinical movement analysis has been traditionally performed in laboratory settings [7]. Much effort has been made to enable movement analysis within ambulatory settings since it has great potential in disease diagnosis and management, thus leading to new research and therapeutic opportunities [7], [8]. Similar research efforts have been dedicated to promote wellness and health in healthy individuals and/or elderly populations. More recently, numerous researchers have emphasized the need for remote sensing systems to assess human movement while donning wearable robotics, e.g., portable exoskeletons as well as stationary rehabilitation exoskeletons, such that optimal feedback and assistance can be provided to the users [4]. Furthermore, there exist countless applications for human movement analysis including the development of novel human-computer interface (HCI) methodologies, integration to smart home and living, and entertainment (e.g., virtual reality games).

Estimation of joint angles is a key component for human movement analysis. Over the past decade, there has been much work focused on developing wearable sensors to facilitate real-time, continuous human movement analysis in free-living conditions. These include approaches using inertial movement unit (IMU) [9], [10], ultrasonic sensors [11], [12], rigid electrogoniometers [13], [14], soft sensors like flex [15], fiber-optical [16], [17], e-textile [18], [19], and liquid metal sensors [20], [21]. However, very few systems are (i) power efficient for sensing and processing of the acquired data, (ii) cost effective, (iii) safe and easy-to-use, (iv) providing flexible form factor to comply with highly dynamic, heterogeneous human body shapes, and (v) supporting the maximum 5 degrees of estimation accuracy that was suggested by the American Medical Association for movement analysis in a clinical context [11], [22].

In this paper, we introduce a novel wearable sensor that addresses the aforementioned limitations of existing wearable sensors for joint estimation and movement analysis. The proposed system employs a retractable string reel and a potentiometer, which computes the amount of skin stretch and finds a correlation to the biomechanical joint movement. A proof-of-concept study involving 9 healthy subjects was conducted to estimate knee joint angles during walking and running. The estimation accuracy of the proposed system was validated against an optoelectronic system, which is the current gold standard method of estimating joint angles.

II. RELATED WORK

Optoelectronic systems are one of the most accurate systems available to estimate human joint angles. These systems employ a combination of infrared cameras to capture the reflective markers positioned on pre-defined anatomical landmarks and create a 3D skeletal model to compute the joint angles. However, optoelectronic systems require an infrastructure (i.e. infrared cameras) in a controlled laboratory setting, and the presence of trained operators to perform the experiment and analyze the data. Thus, optoelectronic systems, despite their accuracy, are not suitable for continuous monitoring of joint movements in an ambulatory setting.

Over the last decade, many studies have introduced various types of wearable sensors to enable joint angle estimation and motion capture capabilities in ambulatory settings. The most prevalent form factor of wearable sensors is the IMU that combines accelerometer, gyroscope, and/or magnetometer [9], [10]. This approach requires extensive signal processing such as Kalman filtering and consistent calibration of the gyroscope for the integration drift, which often requires additional sensing units [20]. The use of gyroscopes, additional sensing units for the calibration, and complex real-time signal processing entails large power consumption. More recently, researchers have introduced an approach that utilizes ultrasonic (ultra-wideband) sensors to estimate angular displacement [11], [12]. The ultrasound transmitter and receiver that are positioned at the extremities of the joint can compute the distance between both sensors, and convert the measurements into joint angles based on biomechanical modeling. This approach provides highly accurate estimation of the joint angles but requires continuous transmission and reception of wireless signals, which again entails large power consumption. The underlying fundamentals of the IMU-based and ultrasonic approaches are similar to the optoelectronic system in the sense that they estimate the joint angles by computing the relative positions of the markers (i.e. sensing units) that are positioned on specific anatomical landmarks.

Another way of estimating joint angles is through incorporating a wearable sensing component that connects the two ends of the joints and directly measures the changes in the angle. An electrogoniometer that uses a rotational potentiometer to estimate the joint angle (e.g. [13], [14]) is a good example that belongs to this category. This is a relatively inexpensive and power-efficient approach to estimate joint angles. However, it often uses hard components (e.g., metal or hard plastic) to connect the limbs, which makes it difficult to comply with the highly dynamic shape of the human body (e.g. obese individuals) and thus, the joint angle estimation is often inaccurate for some groups of people. To overcome this limitation, some studies have utilized more flexible materials to measure the degree of bending of the joint, e.g. flex sensors [15], fiber-optical sensors [16], [17], e-textile sensors [18], [19], and inductive sensors [23]. The flex and fiber-optical sensors are bendable. However, they are not necessarily stretchable and therefore, these sensors have a similar limitation in complying with different body shapes. E-textile and inductive sensors, on the other hand, provide very flexible form factors since they can bend and stretch in different directions. However, they produce a single dimensional output, e.g. resistance for the e-textile and inductance for the inductive sensor, and thus, measurements are often inaccurate. More recently, some researchers proposed to use liquid metal sensors [20], [21] that provide flexible interfaces to the human body and highly accurate joint angle estimation. However, these sensors are relatively expensive as they require sophisticated fabrication processes and more importantly, they may pose a potential safety issue as the materials (e.g. liquid metal) can be toxic.

The wearable sensor proposed in this work, on the other hand, can overcome many of the aforementioned limitations. The proposed soft sensor is flexible, low-cost, power efficient, safe, easy-to-use, and accurate.

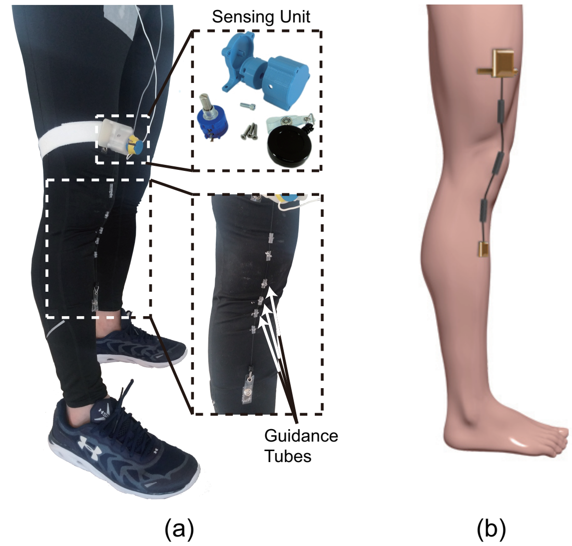

Fig. 1. (a) The prototype of the proposed wearable sensor that utilizes a retractable string sensor (AndyMark Inc., IN, USA [24]), and guidance tubes that appropriately align the string on the human body. (b) The concept design of an integrated system that combines the sensing unit, 9 axis inertial movement units (IMU), a wireless transmission module, an internal memory (micro SD card) module, and a microcontroller.

III. Sensor System

Fig. 1a shows the prototype of the proposed wearable system, which comprises a retractable string sensor and guidance tubes. The retractable string sensor (AndyMark Inc., IN, USA [24]) contains a retractable reel (steel wire), a string, and a potentiometer that measures the amount of rotations of the reel; this sensor is similar to combining a retractable name tag to a potentiometer as shown in Fig. 1a. The sensing unit was placed on the thigh, and the end of the string was attached to the shank. There was a constant tension produced by the retractable reel that pulled the string but this tension was so minimal that participants did not feel it during walking. The string was aligned by the guidance tubes that were sewn on the fabric such that the string did not deviate from its position and trajectory. The potentiometer measured the amount of rotation of the reel during the knee flexion and extension, which was later converted to the change in the length of the string; the length would increase during the flexion and decrease during the extension. The proposed sensor combines the merits of the two broad categories of the wearable sensors that were discussed in Section II. The sensor was designed to compute the distance between the two ends of the joint, but using a soft material (string) that connects the two ends. This sensor works by measuring the amount of skin stretch during joint movements and find its correlation to the biomechanical angular displacement. The proposed sensor is inexpensive (approximately 10 US Dollars for mass production) and flexible. The sensor is power efficient as it incorporates only one potentiometer

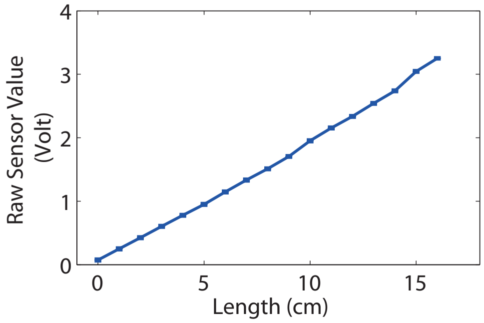

Fig. 2. The empirical relationship between the lengths of the string to the output voltage of the retractable sensor, i.e. potentiometer. The relationship is fairly linear.

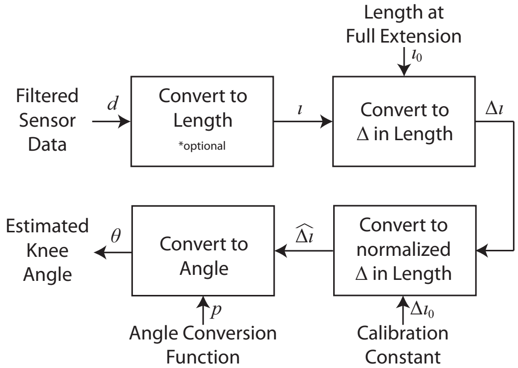

Fig. 3. A schematic representation of the signal processing methods to convert the sensor data to estimated knee angle.

which can operate in a high resistance range ($\sim 0.1M\Omega$). The sensor is safe and easy-to-use (see Section IV-B).

The potentiometer was combined with a Wheatstone bridge circuit and an operational amplifier in order to adjust the operational voltage range (0-3.3V) to the resistance range of the potentiometer. The empirical relationship between the length of the string and the output voltage of the sensing unit is summarized in Fig. 2. The relationship was fairly linear, and the noise level (standard deviation noted by the vertical bars) was insignificant. The plot in Fig. 2 was used to convert the voltage reading (V) to length (cm).

The sensor prototype illustrated in Fig. 1a used an embedded system to collect the sensor data at approximately 50 Hz and converted the analog input voltage to digitized output using an embedded analog-to-digital converter. The output of the embedded system was then transmitted to a personal computer via USB for data storage. Fig. 1b shows the concept design of an integrated system that we are currently developing, which combines the sensing unit, 9 axis IMU, a wireless transmission module, an internal memory (micro SD card) module, and a microcontroller. The system will be placed directly on the human skin, like a Band-Aid, using an adhesive patch.

IV. Methods

A. Data Collection

A total of 9 healthy subjects (a mean age of 32 and a standard deviation of 8) were recruited from the Spaulding Rehabilitation Hospital. The inclusion criteria entailed that subjects could walk and run on a treadmill for up to 3 minutes and were aged between 18 and 80 years of age. Potential subjects were excluded if they had any orthopedic, musculoskeletal, neurological, or any other disorder that resulted in altered gait patterns. An optoelectronic system (Vicon Motion Systems Ltd., Oxford, UK) was used as the gold standard for the measurement of the knee joint angle. Reflective markers were positioned on anatomical landmarks of the lower limbs of subjects based on the plug-in-gait model. Subjects were asked to don the device, and calibrate the system at 0° and 90° extension-flexion knee angles. Then, subjects were asked to walk on a treadmill at 4 km/h and 5 km/h, and run at 6 km/h and 7 km/h for three minutes each. They took a break of approximately 2 minutes between each walking and running trial.

B. Signal Processing

Fig. 3 shows a schematic representation of the signal processing methods used in this work. The raw sensor data was first low-pass filtered at 12 Hz in order to remove any non-human generated noise. The filtered sensor data was then converted to length ι (cm) using the plot in Fig. 2. The length ι was subtracted by the length at full extension $(0°)$ that was obtained from the calibration process. This produced the changes in length $\Delta\iota$ that represented the (absolute) amount of the skin stretched compared to when the legs are fully extended. Then, $\Delta\iota$ was linearly normalized based on the length at 90°, which was again obtained from the calibration process. The normalized (or relative) changes in length $\widehat{\Delta\iota}$ can be computed as

$$\widehat{\Delta\iota} = \Delta\iota/(\iota_{90} - \iota_0) \times 90, \tag{1}$$

where $\Delta\iota = \iota - \iota_0$, ι_0 is the length at $0°$ and ι_{90} is the length at $90°$. This normalized change in length $\widehat{\Delta\iota}$ ensured to normalize the relationship between the knee angle and amount of skin stretch for subjects with different physical characteristics (more specifically, the length and thickness of the lower limbs).

C. Joint-Angle Estimation

The normalized change in length $\widehat{\Delta\iota}$ were then converted to the estimated knee angle using a 3^{rd} order polynomial fitting. The polynomial fitting model was constructed based on the training dataset that contained the knee angle measured by the optoelectronic system and $\widehat{\Delta\iota}$. The knee joint movement requires a complex anatomical model due to non-linear transition of the axis of rotation during flexion and extension [25], [26]; the knee joint does not work as a simple hinge model since the tibia moves non-linearly with respect to the femur. Thus, a 3^{rd} order polynomial fitting was chosen, which best represented the relationship between the ground truth knee angle and $\widehat{\Delta\iota}$. Fig. 4 illustrates the relationships (scatter plots) between the two measurements belonging to one of the participants (Subject #5). Note that the knee

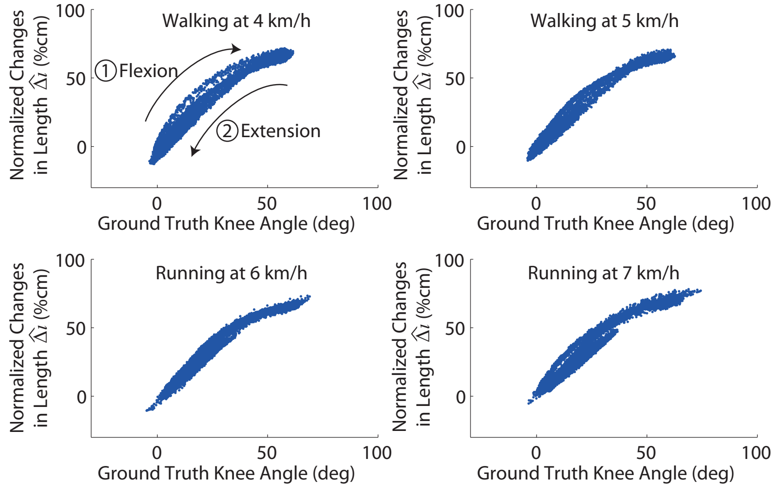

Fig. 4. The scatter plots between the actual knee angle and the normalized changes in length $\widehat{\Delta\iota}$ that belong to one of the participants. The interpretation of the data points according to the flexion and extension cycles of the knee joint is provided in the top left plot.

angle measured by the optoelectronic system was originally sampled at 120 Hz and down-sampled to 50 Hz. The ground truth knee angle and $\widehat{\Delta\iota}$ were manually synchronized.

D. Sensor Calibration

It is noteworthy that the system requires users to measure the sensor readings at $0°$ and $90°$ of knee angle because equation (1) normalizes the length of the skin stretch to the lengths at $0°$ and $90°$. However, because the method first computes the changes in length $\Delta\iota$ and then normalizes it to $\widehat{\Delta\iota}$, the sensor reading at $90°$ needs to be measured only once. This implies that whenever users change the placement of the sensor, they need to measure the sensor reading at $0°$ by simply standing up straight (fully extending the knee), which is a very simple process. During our experiment, we also compared sensor readings when a clinical professional measured the $90°$ with a goniometer and when subjects self-measured the $90°$, and found no significant difference. This supports that the sensor can be easily self-calibrated.

V. Results

Table I summarizes the accuracy of knee joint angles estimated by the proposed system. The results were computed using the leave-one-subject-out cross validation (LOSOCV) technique to validate the systems generalizability. This implies that data of one subject was left-out as a testing dataset, and the polynomial fitting was constructed based on the training data belonging to the rest of the participants. This process was performed iteratively for each subjects. An average root mean square error (RMSE) of $4.51°$ was achieved with the maximum of $6.34°$ and minimum of $3.19°$. Fig. 5 compares the actual knee angle measured by the optoelectronic system and the estimated knee angle of the proposed system that belongs to Subject #5 who produced an error rate (RSME of $4.46°$) that was close to the average RMSE. The results shown in Fig. 5 were obtained during running at 6 km/h.

TABLE I

The accuracy of the knee join angles estimated by the proposed wearable system compared to the gold standard optoelectronic system. The results were computed using the Leave-One-Subject-Out Cross Validation technique.

	RMSE (deg)
Subject #1	6.34
Subject #2	3.60
Subject #3	3.93
Subject #4	3.19
Subject #5	4.46
Subject #6	5.61
Subject #7	5.13
Subject #8	4.82
Subject #9	3.54
Average	4.51

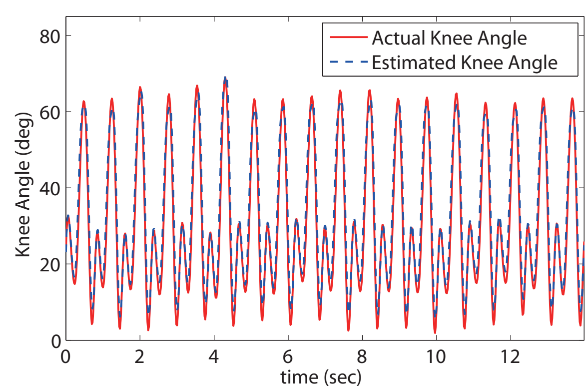

Fig. 5. Plots of the actual and estimated knee angles that belong to one of the participant (Subject #5), who produced an RMSE ($4.46°$) that was close to the average RMSE ($4.51°$)

VI. Discussion and Conclusions

This paper introduced a novel wearable sensor that provides an inexpensive, power efficient, safe, and flexible form factor to measure joint angles of the human body. This soft sensor uses a string to measure changes in the skin stretch that deforms dynamically depending on the physical characteristics and shape of the human body. This proof-of-concept study reported an average RMSE of $4.51°$ when compared to the gold standard optoelectronic system.

Some interesting observations were made. Fig. 4 shows that the relationship between the sensor reading and the actual knee angle created different shapes (or paths) when flexing and extending the knee joint (see the top left plot of Fig. 4). We believe that this was caused by the mechanical property of the sensing component such that when the direction of the shaft of the potentiometer was reversed, i.e. when gait phase changed from flexion to extension, the potentiometer did not respond as fast as it should, resulting in delays in response and causing different paths for extension and flexion. Furthermore, Fig. 5 shows that the estimation errors are higher at lower range of the knee angle, especially during the stance phase when the feet are touching the ground. We believe that this is again due to the delay caused by the potentiometer knob during the stance phase in which the transition between flexion and extension occurs three times (the small bumps in the knee angle as shown in Fig. 5). Future work will investigate the relationship between the tension of the reel and amount of discrepancy of the paths of flexion and extension.

As shown in Fig. 1b, a new integrated system is currently under development. The new system will utilize a rotary encoder instead of a potentiometer since the data processing methodology finds the relationship between the changes in the length of the string to the joint angles; rotary encoder is more adapted to measuring the difference in the amount of rotation rather than the absolute position of the shaft. This will further eliminate the need for measuring the angle at $0°$ for the calibration. Thus, if users measure the angle at $9°$ once, the system can be operated calibration-free.

One limitation of this study is the relatively small number of subjects. However, the results reported in Table I. were computed using the LOSOCV technique, which prevents overfitting and generalizes the use of the proposed technique to a new user. Another limitation is that the optoelectronic system, which was used as the ground truth measure of the knee angle, is known to produce approximately $2°$ of an error rate. Furthermore, the reflective markers were placed on the tight spandex pants (Fig. 1a) and their locations may slightly change during locomotion, which may create some noise to the ground truth data. However, we used very tight-fitting spandex tights and tightly wrapped Coban around the leg to minimize fabric movement, thus we assumed that the impact of these motion artifacts were minimal.

We believe that the new soft sensor introduced in this paper can provide accurate measures of joint angles in an inexpensive and user-friendly manner. The sensor has great potential to be used to monitor knee function in ambulatory settings. We are also in the process of implementing more complex systems that can provide angles of joints having more degrees of freedom using this sensor. This opens up new opportunities for monitoring human subjects in remote settings for potential applications within the fields of mobile health, HCI, smart living, wearable robotics, and entertainment

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