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Wearable Devices for Remote Physical Rehabilitation Using a Fabry-Perot Optical Fiber Sensor: Ankle Joint Kinematic

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ABSTRACT The continuous advances in sensing and telecommunications fields have boosted the development of new technologies towards the improvement of healthcare systems. This drive of knowledge is required to address the rise of life expectancy of an ageing population with increased associated physical impairments, in order to ease the burden on already stressed healthcare systems. Towards such objective, this paper explores the use of a wearable optical fiber based solution for the ankle plantar-dorsi-flexion monitoring, to be used in the evaluation of the progress of physical rehabilitation therapies. The proposed device is a non-invasive, small size and easy to use solution, based on a cost-effective in-line Fabry-Perot interferometer, complemented with new dynamic interrogation techniques that allow the angular monitoring of the ankle-shank joint during gait (walking). The designed and produced wearable solution was calibrated and tested in a laboratory environment, with promising results that prove the accuracy of the wearable device, as it falls within the expected pattern of an ankle plantar-dorsi-flexion movement during gait. The developed system can be used for rehabilitation therapies monitoring, to be integrated in exoskeletons or applied for athletes' performance analysis and optimization, during injury recovery.

INDEX TERMS Ankle plantar-dorsi-flexion, Fabry-Perot interferometers, health monitoring, optical fiber sensors, physical rehabilitation, wearable sensors.

I. INTRODUCTION

Nowadays, thanks to the fast medical technological advances, population life expectancy has been constantly growing, with a predicted value higher than 80 years by 2030 [1]. Although the witnessed technological advances have been contributing towards living longer, now the goal has shifted towards ensuring that we live an autonomous and healthier life, at all ages. Such target requires us to focus on the demands of older citizens' health and the medical technological development needed according to the predicted health problems of older people.

Older age is generally associated with typical physical impairments, especially the spine and lower limb levels, which often require a close supervision from the medical personnel (for physical rehabilitation) or even a strong dependency from care givers to perform the common daily activities. Also, the current sedentary life style of younger citizens is leading to the development of pathologies that impair their normal movement, to the point that they also need physiotherapies and medical aid to regain the natural body physical motion range.

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The accumulation of such physical impairments in the younger and older population results in a heavy burden for the healthcare systems. Hospitals and health care centers are often over populated and there is always a shortage in staff to attend to all patients' needs. Innovative solutions are hence sought after, to decrease the role of the traditional medical provisions in the monitoring of the physical rehabilitation therapies. Such challenges can be met by developing technological wearable solutions towards the aid of rehabilitation therapies, in order to ease the burden on the medical services. Wearable health monitoring solutions are considered the future trends of the healthcare industry. The use of wearable devices, allied to state-of-the-art advances in wireless technologies, promises to provide suitable solutions for remote physical rehabilitation therapies, allowing patients to perform controlled therapeutic exercises from the comfort of their homes, with possible continuous remote supervision by doctors and/or medical staff [2]. Additionally, such solutions can be continuously used, during regular daily routines, to closely monitor the progress of rehabilitation, during usual daily activities. This solution is foreseen to considerably reduce patients' overflow to hospitals and health care centers; thus, reducing costs related to such physiotherapies [3], [4]. Additionally, the storage of the medical data for further analysis, not only helps understand the patient's evolution, but also enables comprehending the health trends within a given community.

When considering the use of wearable devices towards the treatment of debilitated citizens, it should be assured that those wearables do not interfere with the normal daily routine of patients nor limit their range of motion [2]–[4]. Thus, wearables assembly must be carefully thought of, from the production step, up to the implemented sensing mechanisms. The sensors are the key elements of any wearable solution, as they bridge the gap between the physical events and the data that represent those events. Therefore, the accuracy of such physical enablers must be always guaranteed, as they are responsible for the overall performance of the wearable devices, and the data provided for the diagnosis, by the medical staff [2], [3], [5].

One of the most common physical impairments (and a highly restricting one) is related to the normal and healthy locomotion of people. Therefore, in a rehabilitation scenario aiming to achieve the healthy locomotion of patients, it is critical to understand the individual gait pattern (the walking pattern), since the way we walk, and the reason we walk in such way, can be an indicator of several postural disorders, which if timely addressed may prevent complications, and avoid evolving into more severe pathologies.

Commonly, the gait analysis of patients is performed, using video technologies or other expensive platforms based on force resistive sensors [6]–[8]. These technologies are generally only available at medical centers, which also limits the analysis performed, as the patients are conditioned by the intimidating medical environment and its logistics, which are often limited, as well. For the device's response to correctly reflect the patient's limbs movement, they should be worn by patients, without blocking or influencing their range of motion, and with the ability to detect the patients' real movements, during regular daily activities, while ensuring their accuracy, resiliency, and reliability [4], [5]. Unfortunately, the electronic devices have challenges meeting these features.

As an alternative, optical fiber based sensors (OFSs) offer the wearable devices advantages, such as immunity to electromagnetic interferences, biocompatibility, small size (an optical fiber has a diameter of $\sim$125 μm), flexibility, multiplexing ability, and high accuracy. Additionally, since no electricity/current is needed at the measuring point, this technology presents the safety needed to operate in humid or wet environments, as body sweat or water-based physiotherapy routines. Such characteristics make the OFSs a reliable technology to be explored in the design and production of sensing devices, to be further integrated within wearable solutions [9]–[11].

Towards the technological development of wearable devices for gait analysis and the provision of body measurements of rehabilitating patients, in this paper, we propose a novel wearable solution based on OFSs. The proposed wearable solution investigates one of the key segments to monitor during gait (walk), namely the ankle plantar-dorsiflexion. The development of new non-invasive, accurate and reliable solutions to monitor the kinematics of ankle motion is of major importance, not only in the physical rehabilitation fields to help reaching informed decisions and diagnostics, but also towards finding a key role in the sports field, for the analysis of athlete's performance and its optimization [12], [13].

The proposed solution is centered on optical fiber Fabry-Perot interferometric (FPI) based sensors, embedded in a kinesio tape (K-tape), which owing to its biocompatible adherent properties, enables the continuous monitoring of the ankle kinematics parameters during a gait cycle, while preserving the patients' comfort and autonomy. The paper presents the details of the solution design, in addition to the calibration and testing of the produced wearable monitoring solution. The presented results illustrate the high accuracy of the proposed solution.

The methodologies, presented in this paper, are an original approach towards the integration of FPI optical fiber sensors in a wearable structure, for ankle kinematics monitoring. To the best of the authors' knowledge there are no other reports, regarding the use of FPIs for wearable integration and dynamic monitoring, at 1 kHz, of the ankle kinematics. The novel contributions of the paper can be summarized as: i) the development of the low cost monitoring architecture (interrogator) for dynamic monitoring of an FPI modulated signal; and ii) the application of the FPI sensors into a wearable structure of the ankle kinematics monitoring.

The rest of the paper is organized as follows: Section II introduces the kinematic behavior of the ankle, during a gait cycle. In Section III, the design of the proposed sensing mechanism is presented. Section IV is dedicated to the implementation procedures description, and the calibration and testing results. Finally, Section V concludes.

II. ANKLE PLANTAR-DORSI-FLEXION DURING GAIT

The human locomotion is a complex system, where all the interconnected (muscles and joints) move in harmonization to provide a stable and smooth body motion. It is based on the synergy and coordinated movements of the muscles, tendons, flexors, and extensors of the lower and upper limbs, synchronized with the knee, ankle and hip joints rotation on the sagittal plane [7]. Within all the interconnected parts during the gait cycle, the ankle-shank joint has a key role in the process of propelling the body forward, as it establishes the dynamic link between the body and the floor [12], [14]. Fig. 1 presents the typical ankle angular displacement curve, during a single gait cycle [12], [14].

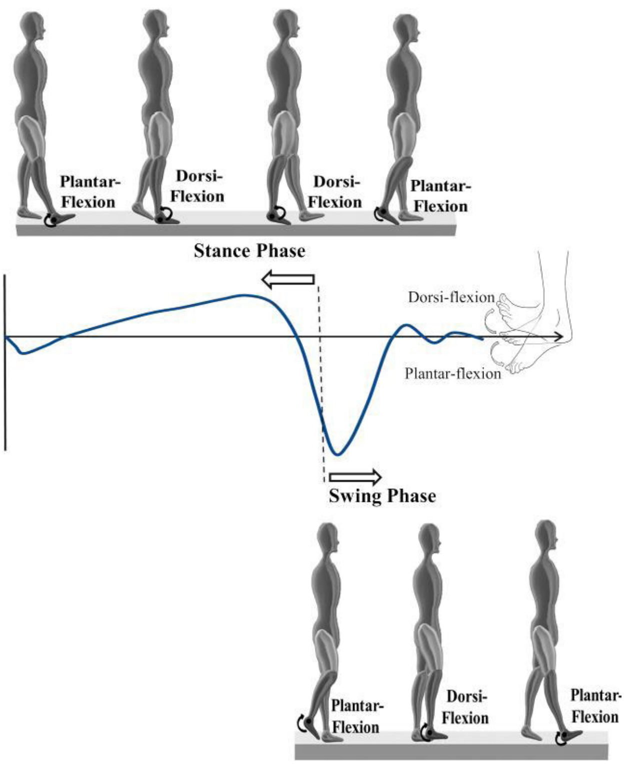

FIGURE 1. Diagram of the ankle dorsi-plantar-flexion and correspondent angle pattern (adapted from [12] and [14]).

The cyclic angular movement of plantar-dorsi-flexion is set between the shank and the foot-long axis, on the sagittal plane, and this is of major importance for the body center of mass movement equilibrium [14]. Considering normal conventions, the foot-shank angle would be preserved at 90°; nevertheless, for the ankle dynamics evaluation, that angle is set as being 0°, while the plantar flexion movement is considered to be the rotation that increases the foot-shank angular distance, and is generally represented by negative values. On the other hand, the angular distance decreases between the foot and the shank, labeled as dorsi-flexion, and is represented by the positive angular movement [12], [14].

A gait cycle can be divided into two main phases: the stance phase, and the swing phase (Fig. 1). The first phase comprises the period of time when the limb has some contact with the floor, while the second characterizes the interval of time during which there is no contact between the limb and the floor. During the stance phase, the ankle main function is to provide a rolling movement on the foot, enabling the leg to progress over the supporting foot. As for the swing phase, the ankle sagittal angular movement is made towards the neutral position, in the preparation for the next gait cycle, when the heel gets in contact with the floor [14].

When considering physical rehabilitation scenarios, the ankle plantar-dorsi-flexion can provide key information for a correct medical diagnoses (e.g. range of motion), and the consequent therapeutic measures to be applied [6], [8].

The relevance of the lower limb kinematics analysis has instigated the research community, towards finding different solutions and techniques for such purpose. Most of the reported and commercially available solutions are based on piezo-resistive, solid-state or stereo optic devices like goniometers or accelerometers [15]. The main disadvantages, associated to the use of electronic-based devices as wearables, are the sensitivity to humidity (e.g. body sweat), and the required bulk, complex or expensive encapsulation materials. Additionally, stereo optic devices cannot be considered, when designing an out of the medical room wearable.

Due to their advantages over the typical electronic devices, OFSs are considered a preferable solution for the proposed application [16]. Nevertheless, its integration, in wearables for the lower limbs kinematics monitoring, has been only recently explored [2], [17], [18]. Few efforts can be found reporting the use of OFSs for the particular case of the ankle kinematics monitoring, and in which cases, the focus is on the use of optical fiber Bragg gratings (FBGs) as the sensing element [18]. Although an FBG is an accurate, reliable and widely explored sensing technology, it can add up to the cost of the sensors production (requires complex and precise recording setups) and the high-cost respective interrogation systems. The use of a cost-effective (both at the production and interrogation levels) FPI optical fiber sensing solution represents a reliable alternative for these applications, as they can outperform the commonly used electronic based goniometers generally used for the ankle kinematics analysis [17], [19]. Promising results, using FPIs for the knee kinematics analysis, were already obtained, with the cost and rate acquisition of the interrogation device being the only disadvantages linked to this sensing technology [19]. In this paper, we take one step forward in the use of such cost-effective interrogation device, by proposing - for the first time - the use of FPI sensors in the dynamic monitoring of ankle plantar-dorsi-flexion.

III. PROPOSED SENSING ARCHITECTURE

In this section, we detail the design and methodology, used for the assembly of the wearable sensing device for ankle kinematic monitoring.

A. FPI SENSORS PRODUCTION & WORKING BASICS

FPI-based sensor is a widely used sensing technology, due to their accuracy and efficiency, in addition to their low-cost production techniques [20]. The sensor production follows a cost-effective technique and was first described in [21]. It consists in the recycling of the optical fiber, previously damaged by the fiber fuse effect in a sequence of steps that lead to the final formation of an in-line FPI micro-cavity, as illustrated in Fig. 2.

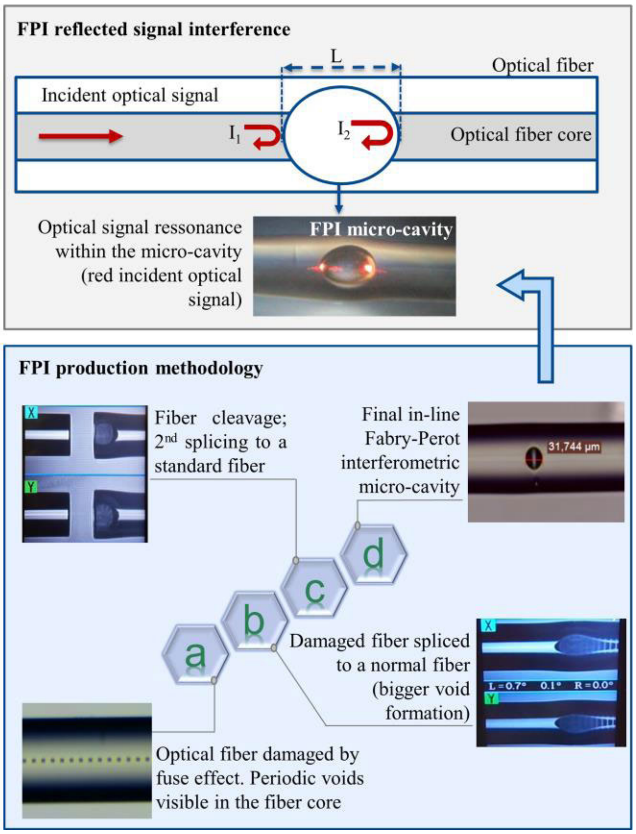

FIGURE 2. Schematic representation of the FPI micro-cavity based sensor head production and working principle.

The damaged optical fiber presents in its core a trail of periodically spaced voids, which prevent the transmission of an optical signal, rendering it unfit for telecommunications purposes (Fig. 2a). When this fiber is spliced to a standard single mode fiber (SMF), the initial voids induce the formation of a bigger void, as depicted in Fig. 2b. For the formation of the single in-line FPI micro-cavity, the fiber is cleaved and spliced again to an SMF fiber, as depicted in Fig. 2c. The obtained final micro-cavity is illustrated in Fig. 2d.

The interference reflection optical spectrum, created due to the resonant FPI micro-cavity, has an intensity profile (I), dependent on the micro-cavity length (L), and the refractive index (n), of the voids constituent material. Equations (1) and (2) describe such dependency:

$$I = I_1 + I_2 + cos\,(\delta)\,\sqrt{I_1 I_2} \tag{1}$$

where I_1 and I_2 are the intensities of the reflected optical signal in the two mirrored surfaces of the FPI, and δ represents the round trip optical phase difference between two adjacent signals of wavelength, λ, at a normal angle of incidence, given by:

$$\delta = \frac{4\pi nL}{\lambda} \tag{2}$$

Therefore, any change in the micro-cavity length/volume or refractive index induces a change in the reflected interference. In Fig. 3, the optical interference spectrum, obtained for the produced micro-cavity, is illustrated in two different situations, with and without an applied external strain. The fiber elongation, resulting from the applied strain, results in the increase of the micro-cavity length, from L to $L + \Delta L$, leading to a wavelength shift ($\Delta\lambda$) in the FPI optical transfer function. The wavelength shift, induced by the applied strain, represents the FPI sensor feature exploited for the production of the wearable optical fiber sensing device.

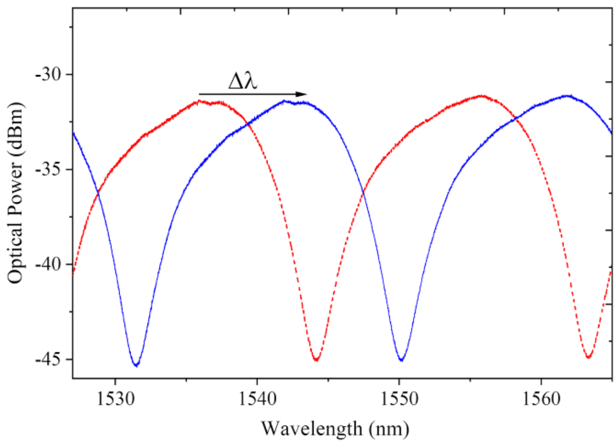

FIGURE 3. FPI spectral displacement due to the optical fiber elongation (Dashed red line corresponds to the signal without elongation and blue continue line to the signal with an applied elongation).

B. 3-D PRINTED WEARABLE FPI SENSOR DESIGN

In its bare condition, the silica optical fiber is a fragile thin glass wire (~125 μm of diameter), which may break due to an external actuation. Nevertheless, it can become more robust, by means of encapsulation, allowing its use in harsh environments or challenging applications. In the case of this particular biomedical application, the optical fiber sensor encapsulation not only provides the necessary robustness, but also enables its re-use.

Fig. 4 depicts the encapsulation process designed for the FPI optical fiber sensor and its final integration for ankle kinematics analysis. First, the bare optical fiber, containing the sensor head, was placed in a polylactic acid (PLA) 3D printed mold (~5 cm) and fixed on both extremities outside of the mold, to ensure a slight strain and straight horizontal alignment of the optical fiber (Fig. 4a). Next, the mold was filled with a thermosetting resin (LiquidLens), and left to cure for 24 hours. This resin has a strong adherence

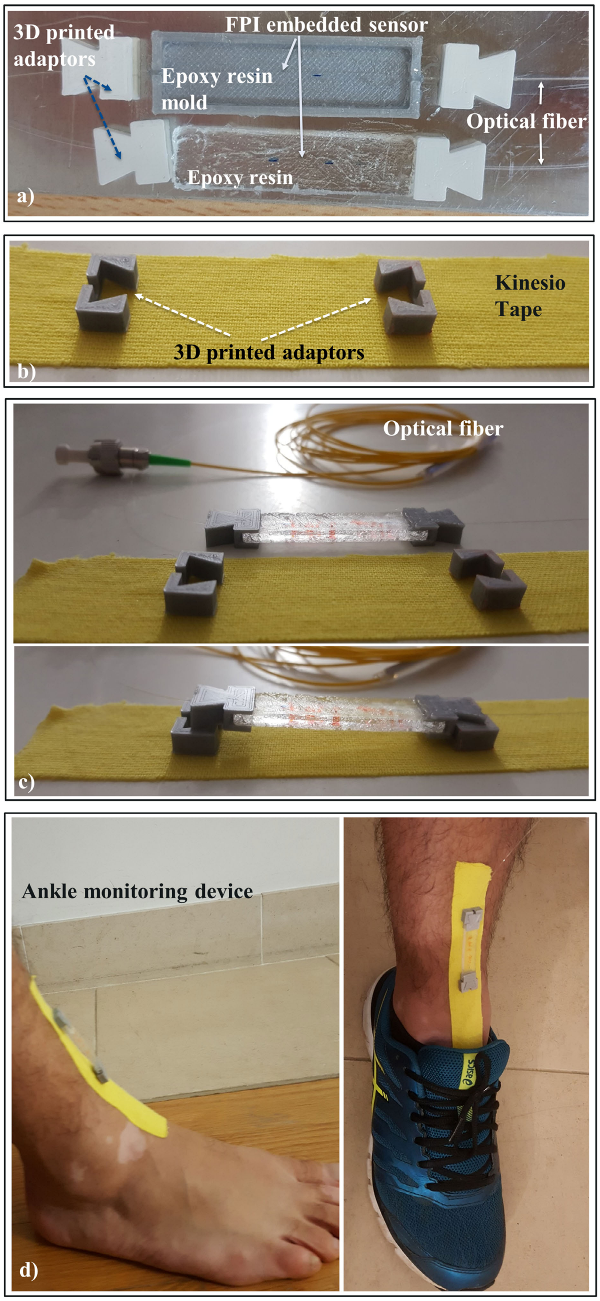

FIGURE 4. 3-D printed encapsulation structure for the plantar-dorsi-flexion monitoring and its application.

to the optical fiber, guaranteeing that any deformation applied on it will also be reflected on the sensor. After unmold, the extremities of the resulting resin sensing structure were firmly attached to two PLA 3D printed adaptors. These adaptors fit in the 3D printed sockets (Fig. 4b), allowing the re-use of the sensors for several applications and their easy-to assemble by hooking/unhooking at the 3D printed socket.

To make the developed sensing device wearable, the described structure was integrated within a kinesio tape (K-tape), which is the element that will be in contact with the limb. The K-tape is a hypoallergenic and wearable elastic tape, frequently used in the treatment of muscular conditions. Its composition (elastic cotton and an acrylic adhesive side for a strong adherence to the skin) provides stability to muscles and joints, without jeopardizing the motion's range and the normal limbs' movement [22]. The 3D-printed sockets were integrated within the K-tape, by a self-cured strong adhesion acrylate glue, as shown in Fig. 4b. The sensing structure can then be hooked onto the sockets as illustrated in Fig. 4d. For the ankle kinematic monitoring, the K-tape, containing the sockets, needs to be placed along the lower limb shank down till the beginning of the foot (Fig. 4d). The sensing structure is then hooked up into the sockets.

As the sensing architecture is based on the strain induced by the stretching of the K-tape, it is important to choose a mounting location that does not suffer relevant muscle tension during gait, which could overlap with the ankle kinematics [3], [13]. Also, a relevant consideration, that needs to be taken into account for the design of any wearable sensor, is the comfort of the user and the least impact possible on the user's movements [2], [3], [13].

With the ankle sagittal movement into a plantar-dorsi-flexion position, the K-tape stretches and compresses, accordingly. As the K-tape stretches with the plantar-flexion, it induces a proportional tension in the epoxy resin structure, translated by the spectral wavelength shift ($\Delta\lambda$) of the FPI modulated optical signal, as previously illustrated in Fig. 3. A spectral wavelength shift in the opposite direction is obtained with the compression of the K-tape within the dorsi-flexion movement.

It should be noted that this is a comfortable, non-invasive and biocompatible wearable solution. Since the sensor is in direct contact with the limb part, the measurement errors, often associated to sensor displacement during the walking or the performance of the rehabilitation exercises, are reduced.

The described sensing architecture was characterized to temperature, deformation and afterwards tested during a volunteer's normal gait.

C. ACQUISITION ARCHITECTURE

FPI sensing technology has often been only used for static monitoring of parameters, such as strain, temperature and refractive index, due to the available interrogation/analysis techniques [20]. Such interrogation techniques, on the frequency domain, require a full spectral acquisition, which renders the commonly used equipment expensive, or unreliable for dynamic acquisitions. Such constraints have been slowing the adoption of FPI sensors in biomedical applications, such as the monitoring of kinematic parameters during gait. Recently, we have described an alternative interrogation technique, based on optical power intensity analysis that renders FPI sensors suitable for the dynamic monitoring of parameters at a high rate frequency, up to 1 kHz as reported [23], or depending on the photodiode specifications. Similarly, the interrogation technique, used in this work, is based on the analysis of the integrated optical power, resulting from the convolution between a narrow bandwidth optical signal (laser source) and the FPI spectral transfer function. As the FPI spectral transfer function is modulated by the deformation (stretching and compression of the K-tape), the initial optical power undergoes a variation, as illustrated in Fig. 5.

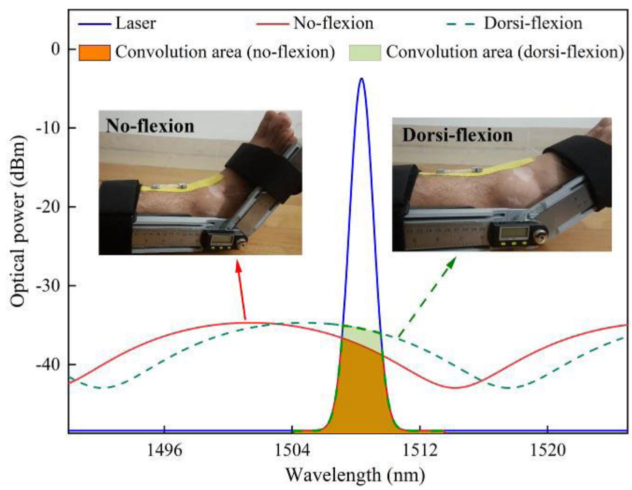

FIGURE 5. Illustration of the sensing principle of the proposed solution.

With the ankle movement into a plantar-flexion position, the FPI is tensed, and the induced red spectral shift increases the reflected optical power, resulting from the laser and the spectral convolution integrated area. On the other hand, a compression of the K-tape in a dorsi-flexion movement induces a blue shift on the optical spectrum, and a consequent decrease in the reflected optical power. The overall architecture, comprising the interrogation solution described in [23], renders the FPI sensing device a cost-effective solution for dynamic rate acquisitions. This innovation in wearable sensing architecture paves the way for FPI sensors to be further exploited in more diverse fields of applications. Up till now, the application of FPI sensing was limited, due to the high costs required by the interrogation devices, for high-rate acquisition applications, as required by applications, such as gait analysis.

The apparatus, used for the overall architecture and depicted in Fig. 6, comprises a narrow band optical source at ~1508 nm (FOL1405RTD-657-1509, Fitel, Tokyo, Japan), an optical circulator, the sensing device and a photodetector (Model D400FC, Thorlabs, Newton, Nova Jersey, EUA) for the optical power analysis, with an acquisition rate that can reach up to 1 kHz, which considering the envisioned application, allows the observation of small drifts, such as peak activities, registered at the muscles location.

The photodetector is connected to the computer, by means of a digital to analog converter (USB6008, National Instruments, Austin, Texas, EUA). It should be noted that, within the lab, the overall used apparatus is still bulky; nevertheless, similar components solutions (lasers/diodes), with a considerable reduced size, are available in the market at affordable prices.

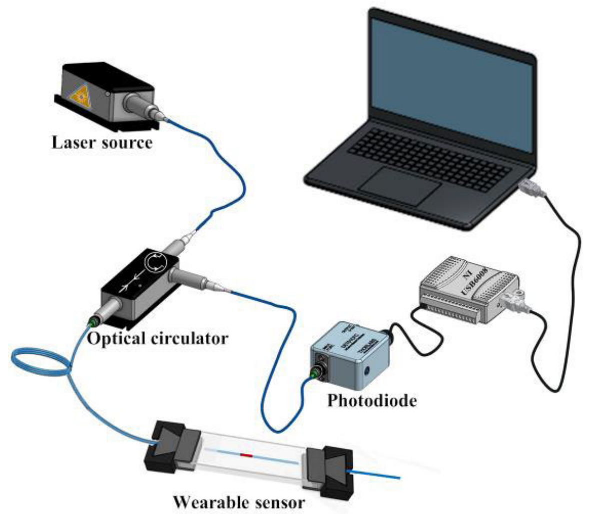

FIGURE 6. Interrogation setup for FPI sensors, based on the convolution technique between a narrow band optical signal and the FPI spectrum.

IV. IMPLEMENTATION AND RESULTS

Prior to its application in the monitoring of the ankle kinematic parameters during a normal gait, the sensor's response was characterized to temperature and deformation variations. The latter allows to convert the optical power into angular variation of the ankle joint. The following sections focus on the characterization methodology and testing of the proposed architecture.

A. TEMPERATURE AND DEFORMATION CARACTERIZATION

The sensor was thermally characterized, using a climatic chamber (CH340, Angelantoni Industrie, Italy). The temperature was gradually increased from 15.0 to 50.0 °C, with incremental steps of 5.0 °C. The reflected transfer function of the FPI was registered for each temperature level, after a stabilization period of 30 min. An average temperature sensitivity coefficient of 0.054 ± 0.002 nm/°C was achieved. This sensitivity is mainly due to the resin encapsulation and its physical response to temperature changes, since the previous studies showed that an FPI without encapsulation has a lower thermal sensitivity (0.050 ± 0.001 pm/°C) [21]. Nevertheless, it should be noted that, as the optical fiber sensor structure itself is not in contact with the body, no correlation is expected between the acquired data and the body temperature. Also, the results, presented here, were acquired in a controlled temperature environment, and therefore were not affected by temperature variations. Additionally, it was given some stabilization time between the moment that the sensor is placed and the beginning of the tests. As for the applications, where a temperature variation is expected (ex. walking between different rooms, or out of the lab testing), some temperature compensation techniques, similar to the ones described in [24] can be integrated within the proposed architecture. For the calibration of the FPI based sensor to ankle angular movements, a customized digital goniometer was used, as shown in the inset of Fig. 7.

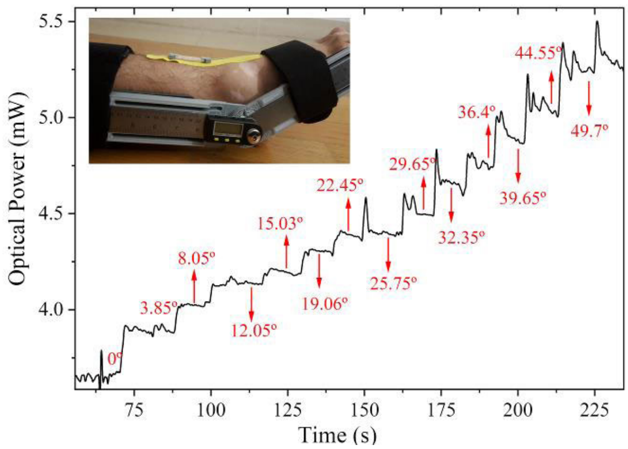

FIGURE 7. Dependence of the optical power with the applied angle. Each step corresponds to a registered angle. The data was acquired in continuum with an acquisition rate of 1 kHz. The red numbers correspond to the angle registered by the goniometer.

After the sensor is placed along the shank, the foot was positioned in an immobilized, yet conformable position. Then, one brace of the goniometer is firmly attached to the foot, while the other arm is placed along the shank, so that the movement of the ankle joint is replicated in the goniometer. For the calibration purpose, the foot was moved from the maximum dorsi-flexion up to the maximum plantar-flexion, in an increment of ~ 3 degrees, stabilizing the foot for a few seconds in each position. Considering a rehabilitation scenario, this procedure should be made prior to each session, given that the K-tape is removed in between sessions. If the K-tape is kept in place in between therapy sessions, the first calibration should stand, as the anchorage points for the sensor do not change.

The angle, at which the foot stops (given by the goniometer), was registered, while at the same time the photodetector was continually registering the optical power variation. Fig. 7 displays the optical power variation, as a function of time, for each angle increment applied. In the same figure, the angle values, registered by the goniometer, are clearly depicted.

In Fig. 8, the relation, between the ankle joint sagittal rotation angle and the optical power registered by the proposed sensing architecture, is presented, for two arbitrary tests.

The hysteresis effect, for this type of FPI sensors, was previously evaluated for a bare fiber [21], and also considering its encapsulation in an epoxy resin structure, similar to the one used for this specific application [23], where a maximum hysteresis of 0.070 nm was reported. These previously reported results show that the hysteresis values associated to this sensing units are not relevant in the setup, presented here.

The achieved sensibilities were 0.0296 ± 0.001 mW/° for calibration 1 and 0.0279 ± 0.001 mW/° for calibration 2. Although the sensitivities appear to be different, its uncertainty falls within the predicted error range, which

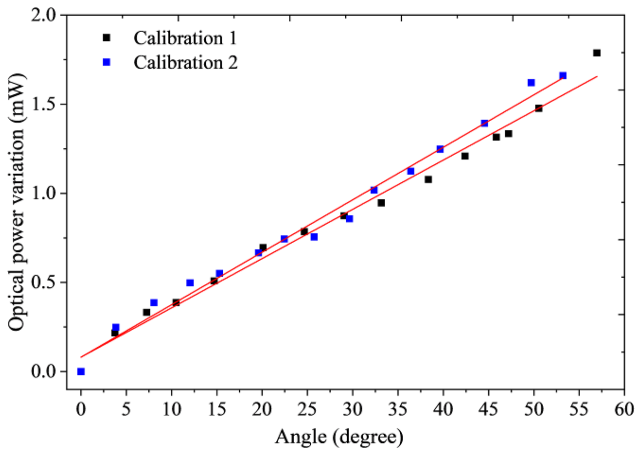

FIGURE 8. Optical power variation as function of the angular displacement. Points are experimental data and the lines correspond to the linear fits.

can be due to the optical source power oscillations. These oscillations can be corrected by adding a signal splitter (ex. 10% / 90%) and a second photodiode for the source optical power continuous monitoring. Based on the two calibrations presented for the range of tested values, and from the linear fits performed (R^2 ~0.997) to the experimental data, an average sensitivity (S_{FPI}) of 0.029 ± 0.001 mW/degree was achieved. This is the value to be used to convert the optical power variations (ΔOP), detected at the photodiode, into the corresponding angles, according to the relation:

$$Angle = S_{FPI} \times \Delta OP \tag{3}$$

B. KINEMATIC ANKLE ANALYSIS DURING GAIT

After the angular calibration, the sensor device was kept intact, and the volunteer was asked to walk at a normal pace in a straight line. In Fig. 9, the angular variations are plotted, which are acquired during the requested exercise, when the volunteer walked in a straight direction, turned around, and walked back to the initial position.

The turning point is clearly visible in the graph. Additionally, the different gait cycles (GC) in each direction are easily identifiable. The similar pattern observed during the registered gait cycles reflect the plantar dorsi-flexion of the ankle, as theoretically expected [12], [14], [25]. Also, from the plotted data, it is possible to infer the duration of each gait cycle (~1 second).

Fig. 10 presents a deeper analysis of the ankle plantar-dorsi-flexion kinematics for the first gait cycle (GC1). At the beginning of the gait cycle, when the heel strikes the floor, the ankle is at a neutral position, evolving for a slight plantar-flexion in the initial stance phase (please refer to Fig.1 for the theoretical ankle kinematic gait pattern comparison). With the progress of the gait cycle, at the foot-flat moment, the shank starts drifting from a plantar flexion into a dorsi-flexion position, with the heel and the forefoot placed in full contact with the floor, ready to advance the body forward.

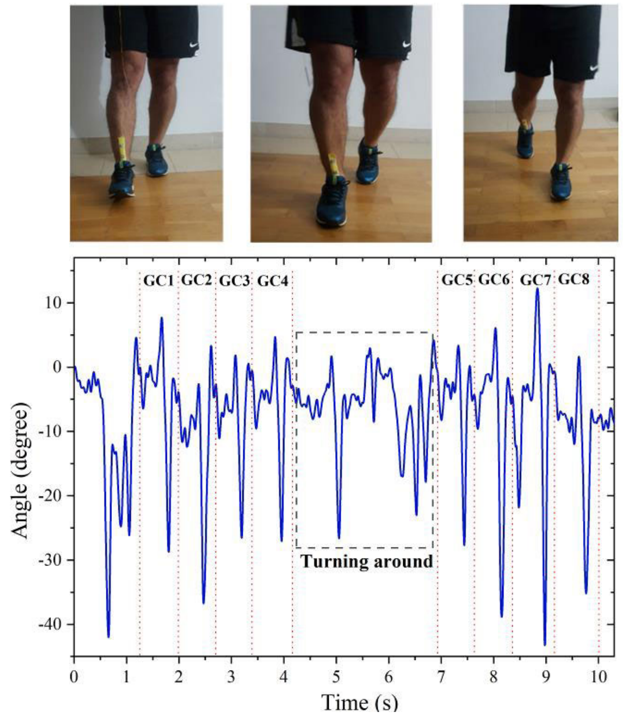

FIGURE 9. Ankle angular variation during a normal gait for a total of 8 gait cycles (GC).

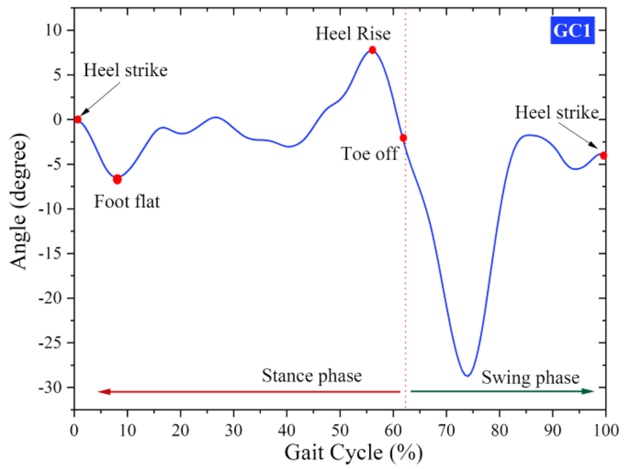

FIGURE 10. Detailed analysis for the GC1 of the ankle angular kinematics.

At the middle of the stance phase, the leg and foot muscles and extensors (like the extensors digitorum brevis, EDB) are activated towards the control of the plantar-dorsi-flexion transition, which is also perceptible in the acquired data, by the slight peak increase between the 10% and 20% of the gait cycle. This data is sensed by the device, because the K-tape is glued on the foot surface up to the EDB location. Therefore, upon its peak activity, the K-tape is additionally stretched to the expected movement of the ankle angular rotation [26]. With the progress of the gait cycle, as the heel leaves the floor, the ankle rapidly shifts from a dorsi-flexion into a plantar-flexion position. After the toe off moment, when the swing phase is initiated (see Fig.1), the ankle moves from the maximum plantar flexion, into a slight dorsi-flexion, in preparation for the next heel strike of the flowing gait cycle, with the ankle at a neutral position.

We would like to emphasize that, the work, presented in this paper, is related to the preliminary implementation of the sensor for ankle kinematic monitoring; therefore, we only considered the data acquired from a volunteer without pathologies, to test the feasibility of the proposed architecture for the considered application. Deviation from the initial conditions of the K-tape mounting (like the ankle swelling, tape deviation in between therapy sessions), can be mitigated through the initial calibration parameters and the software associated to the data acquisition. Based on the calibration parameters, three reference values of strain can be initially set for maximum dorsi-flexion, zero degree, and maximum plantar-flexion. Such references give us the amplitude of strains, expected for each user. Any deviation on the reference strains, especially the one associated to zero degrees, sets an alert to confirm the device position, and if needed to reset it to a new calibration.

V. CONCLUSION

The ageing of the population and the current sedentary life style of the younger generations is leading to a crescent of physical impairments. Such pathologies require a constant monitoring of its evolution and the physiotherapy progress, which can be achieved through wearable sensing devices. This paper proposes a new technology to provide more accurate activity monitoring measurements for physical rehabilitation, targeting wellness and sport applications. In this work, we explored the use of a new wearable optical fiber based sensing technology to continuously monitor the ankle kinematics parameters, during gait. The sensing device is based on an optical fiber in-line FPI, integrated with a new dynamic interrogation technique that guarantees the resolution and accuracy of the sensors, necessary for this type of applications. The proposed innovative wearable solution was shown to be suitable for measuring the angular plantar-dorsi-flexion of the ankle, during walking without jeopardizing the user's autonomy nor limiting their range of movement. The device was tested during walking, in a laboratory environment, providing accurate results in accordance with the theoretical expected ones.

Device optimization will be the subject of future work, which will comprise the inclusion of machine learning adaptive training algorithms, to understand the sensor's response based on its placement location, and automate a self-calibration considering some reference points (like zero, maximum plantar, and maximum dorsiflexion positions).

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