Development of an Optical Fiber-Based Insole for Gait Parameters Detection in Rehabilitation Processes in the Population with Lower Limb Disabilities

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Final project for specialization in rehabilitation engineering

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ABSTRACT

Introduction: As several diseases are reflected in the human foot as plantar pressures, divers gait assessment technologies have been developed. Some of them include force plataforms, IMUs (inertial measurement units) and instrumented insoles. Instrumented insoles have advantages over the other mentioned technologies, for instance, they do not limit the gait assessment to few steps and, they do not require a repetitive process of calibration. Different sensors have been used to develop these instrumented insoles, some examples are capacitive sensors, FSR, piezoresistive sensors and optical sensors. In literature optical sensors have been identified as an attractive option on account of their mechanical properties and the electromagnetic noise immunity. Taking into account the state of the art in this work, an instrumented insole based on polymeric optical fiber sensors is developed and characterized.

Objective: The objective of the present work is to develop an instrumented insole, with a POF sensor and, an acquisition and transmission module for gait assessment.

Methodology: To develop the instrumented insole the project was divided in three main components which are the insole, the sensor and the acquisition and transmission module. The sensor was made with a Polymeric optical fiber and the operation principle of the sensor was developed based on a multiplexing technique. The described components were assembled and a characterization protocol was implemented. This protocol contemplated intervals of time to measure the sensor response to different loads and time intervals for the sensor recovery.

Results: In this section the characterization protocol outcomes were showed. The response to loads of the four sensitive zones were no-homogeneous among them. The majority of the sensitive zones presented s linear behavior with some exception. On the other hand, some of the sensitive zones presented R^2 values lower than $0{,}8$ and others reached R^2 values higher than $0{,}9$ in their linear trend.

Discussion: Several factors affected the optical fiber sensors response whereby they have to be consider, otherwise, this factors would affect the developed measuring system, resulting in non-repeatability and in linear response to load decreasing as could be observed in the characterization outcomes.

Conclusions: An instrumented insole and its acquisition and transmission modules were developed using the microcontroller of the FRDM KL25Z board; additionally an optical fiber coupling board was use to integrate the light sources and the optical fiber for the sensor to work correctly. The operation principle of the instrumented insole is based on a multiplexing technique.

1. INTRODUCTION

Different pathologies like Charcot-Marie-Tooth (CMT) disease [1], Hansen disease [2], rheumatoid arthritis [3] and COPD (Chronic Obstructive Pulmonary Disease) [4] are reflected in the human foot as plantar pathologies [5]. For that reason, advances in pathologies detection are related to the gait biomechanics' analysis [6]. Gait phase detection is part of the gait biomechanics' analysis and is relevant to rehabilitation processes, early detection, motor impairments from diverse origins assessment, and prevention [7]. At present, there are various options to evaluate gait. Technologies that allow measuring biomechanics characteristics related to gait are considered kinematic methods [8], and plantar pressure measurements methods [9] to which anomalies in biomechanics associated with musculoskeletal illnesses in different zones as knee, back, ankle, and different pathological origins can be characterized [6]. Some of these technologies are force platforms, instrumented treadmills and insoles, multi-camera systems and electromyography [7] [10]; those devices measure the three-dimensional force and the markers' positions accurately [10]. Nevertheless, what is wanted from these technologies is that tests' setups do not modify gait natural patterns, that it allows assessment in different contexts and grounds, and to be low-cost [11] [7]. However, each one of the technologies mentioned above presents disadvantages. In the case of force platforms, they segment the gait by assessing just one step characteristics [7]; instrumented treadmills disrupt the normal gait parameters [10]. Regarding instrumented insoles, sensors used have limited durability and measurements are impaired by the shoe inner contour [9], and according to the type of sensor that is used, the measurement can be affected by hysteresis, non-linear response, and sensitivity to temperature, among other factors [12]; but their accuracy surpasses the one obtained with IMU (inertial measurement unit) in determining the vertical ground reaction force $GRF_v(t)$, anterior-posterior ground reaction force, $GRF_{ap}(t)$ and medial-lateral ground reaction force $GRF_{ml}(t)$ [9]. Additionally, they allow measurements outside laboratories and in real environments in which the life of persons is developing [13].

For instrumented insoles, Morère et al. raise that the requirements are linearity, low hysteresis, sustained pressure of $1900kPa$ (range of pressure reached in the foot during gait cycles [14]), sampling frequency minimum of 200Hz, low-temperature sensitivity and repeatability [12]. Leal-Junior et al. highlight insole lightweight, compactness, sensors distribution in the highest pressure areas, and sensors' small size [14]. In light of the mentioned requirements, different sensor types have been posed for insoles design, among which are Force Sensing Resistors (FSR), capacitive, piezoresistive [12], piezoelectrics, textile, conductive rubber, optical fiber, and air pressure sensors, among others [15]. These sensors' functional principles and characteristics determine the number of sensors required for the insole development, the necessity of other sensors like gyroscopes or accelerometers, and the developed instrumented insole properties [15].

With regard to FSRs or Force Sensing Resistors, these are sensors formed by two membranes and a layer in the middle; in the inferior layer, there is an active zone that changes its resistance value responding to the applied force on the top layer [16] [17]. With the force increment, the sensor's resistance decreases in an exponential form [18]. FSRs are highly used for compact wearable systems development for plantar pressure detection [5]. Jord et al. developed an instrumented insole with FSRs to evaluate plantar pressure static component; the results were compared with results from commercial devices and comparison relativity coefficients between $R^2 = 0{,}9335$, and $R^2 = 0{,}9963$ were obtain [19]; though, from the characterization protocol of the insole with 9 FSRs developed by Rescio et al. resulted in a non-homogeneous behavior of the FSRs [20].

In the case of piezoresistive sensors, these are made of type P and N materials. P material is merged into N material, and they present a resistance variation when there is a compression or stress force upon them [21]. These sensors allow plantar pressure measurements with ease electronic and with less noise affecting compared with capacitive or piezoelectric sensors [22]. Commercial references of piezoresistive sensors allows measurements of a huge pressure range that can achieve $6{,}24MPa$ [23]. On the other hand, some research works that are about the development of sensors. One of the most used materials is the graphen, Lou et al. and Tan et al. used it and they obtained a sensitivity range from 0 to $50KPa$ and the same maximum pressure value of $800KPa$ [24].

Furthermore, capacitive sensors' functional principle is described by their capacitance, ϵ_0 the vacuum permittivity, ϵ_r the relative dielectric permittivity, A the electrodes (films) area and d the distance between the electrodes [25]. In the majority of cases pressure capacitive sensors measure pressure based on the capacitance variation due to an external force effect in the separation distance d [25]. Yet in their research work, Motha et al. other option and raise varying the ϵ_r parameter for determining pressure changes with higher sensitivity. Nevertheless, their proposal requires nanoparticles to use and PDMS (polydimethylsiloxane) for replacing the surrounding air, this is done by encapsulating the sensors in PDMS [26]. Some commercial technologies for gait assessing like The Pedar system (Novel® GmbH, Munich, Germany) use these type of sensors [27]. Some of these sensors' advantages in comparison with others are their low sensitivity to temperature and humidity and good repeatability [28]. Furthermore, a disadvantage is their logarithmic response, even though, Aqueveque et al. affirm that in their proposition this occurs just between 0 to $500kPa$ and that the area of interest is from 500 to $1200KPa$ in which the sensor has good sensitivity and a linear response [29].

Finally, the optoelectronic technologies consist of an optic mechanism where there is a light-emitting phase, a light receiver, and a transmission medium that is occluded by the externally applied pressure [27]. In the fiber-based sensors, the objective is to transmit information by one light parameter modulation of the light that goes through the fiber. That parameter can be the intensity, wavelength, phase modulation, spectrum, among others [30]. In the same way, there are alternatives for measuring the modulated light parameter. The FBG (Fiber Bragg grating) and the intensity methods are the most used [31]. Optic fiber sensors (OFS) are characterized by high sensitivity, efficiency, and resolution [32], moreover, the electromagnetic noise immunity [33]. On the other hand, these sensors despite being sensible to factors like temperature and humidity, they do not have huge influences on the gait cycle assessment because the gait cycle is faster than these sensors' time response to the aforementioned [34]. Regarding the measuring methods, the FBG principle consists of measuring the Bragg wavelength variation due to an external force applied in a fiber with a grating in its core [35]. When there is pressure on the fiber, the diffraction grating period Λ changes. Therefore, the Bragg wavelength λ_B changes [35]. This method requires an expensive interrogator that makes the measurement system less compact [33] and with a low sampling frequency [36]. On the other hand, the intensity variation method is less complex to implement and less expensive than FBG's method [33]. For the intensity variation method, the fiber has to be adapted by making cuts from the jacket to the core of the fiber [37], these cuts are called sensitive zones [38]. Hence, when pressure is applied on the sensitive zones, the intensity at the end of the fiber is reduced, and photodetectors are used for sensing the intensity [37]. However, this method has less multiplexing capabilities than other methods, but a multiplexing technique for intensity variation technology was developed in "Polymer Optical Fiber-Based Integrated Instrumentation in a Robot-Assisted Rehabilitation Smart Environment: A Proof of Concept" [36]. Moreover, OFSs' characteristics are determined by fiber material; SOFs (Silica Optical Fiber sensors) present fewer transition losses [39], but there is the risk of getting fracture due to the high pressures [35]. And POFs (Polymeric Optical Fiber sensors) are less affected by temperature and their sensitivity and repeatability are higher [39], they have a lower Young's Module meaning that have more flexibility and therefore more pressure sensitivity [14], and yet the polymeric nature of the POF can represent non-linearity, in some works a linear response has been found [36], [38].

This paper presents the development of an instrumented insole based on POF sensors with four sensitive zones, the building process of the insole, the adaptation (structural changes) of the fiber and the characterization of the sensitive zones.

2. OBJECTIVES

2.1. General objective

To develop an instrumented insole with POF sensors, design an acquisition module, a transmission module and a storage module to build a system that allows the reception and storing of pressure signals and determine the insole scope for reliable detection and timely follow-up of gait events.

2.2. Specific objectives

1. To establish the architecture of the instrumented insole and the Acquisition and transmission modules.
2. To implement a multiplexing technique to generate pressure signals in the acquisition module.
3. To develop an algorithm in a processing unity for the data acquisition and storing.
4. To characterize the developed instrumented insole (the four sensitive zones of the POF fiber) and its functioning.

3. METHODOLOGY

This section presents the process of the insole development, a description of the operation principle of the lateral sections and their design; Additionally there is a description of the acquisition and transmission modules composition

3.1. Operation Principle and Sensors Design

3.1.1. Operation principle and sensitive zones elaboration

A Polymethyl-Methacrylate resin optical fiber (MIKROE-1473) with a typical core diameter of $980\mu m$ was used to measure four white light LEDs (WHITE 3528 SMD LED) optical power through the multiplexing technique. The multiplexing technique consists of having a source of light in front of each sensitive zone [40]. It works by turning on one LED at a time following a sequence with an established frequency. It requires two photodetectors, one at each end of the fiber and a processing unity that makes the signal acquisition and controls the logic outputs to turn on the LEDs [40]. In this case a LED turning on routine was programmed in a development board (FRDM KL25Z) at a frequency of 30 Hz. Only one LED at a time was turned on using the mentioned frequency.

The sensitive zones are the pressure sensors that will transmit the light intensity variation when pressure is applied to them. Some important parameters to consider in the elaboration process of the sensitive zones (also called lateral sections) are their length and depth [41]; for that reason, to elaborate the lateral sections, the fiber was introduced into a structure with a slot, this structure was used as a guide to making the sensitive zones as homogeneous as possible. Then with the aid of a motor tool, the fiber was scraped further on the fiber core.

3.1.2. Insole Instrumentation and Setup

The complete pressure measurement system includes an insole comprised of two pieces, the base, and the cap. The insole design was developed in SOLIDWORKS (software CAD) and it was printed through the software Simplify3D where all the printing parameters were established. Both the base and the cap of the insole were printed in TPU (Thermoplastic Polyurethane); these pieces were designed based on shoe size 39 EU.

Once the two pieces of the insole were elaborated, the fiber placement in the insole proceeded; for fiber positioning and fixation there was a narrow path in the insole base where there were four circular spaces, one for each sensitive zone. The circular spaces' positions were chosen to be in the higher plantar pressure spots. In the bottom layer of the insole cap, there are matching parts to the circular spaces in the base. These parts have a slot in the middle for placing the LEDs and let the light sources be in front of the lateral sections. That configuration and the lateral sections numbering can be appreciated in Fig. 1.

To add robustness, the spaces in the insole base that are around the fiber path were filled with Ecoflex 00-30 which is a silicone based in two liquid components that become solid when they get mixed. The rubber silicone was prepared by mixing its two components using a ratio of 1:1 ($60g$ of each component); then the whole content is poured into a syringe to eliminate air bubbles by creating vacuum different times for about five

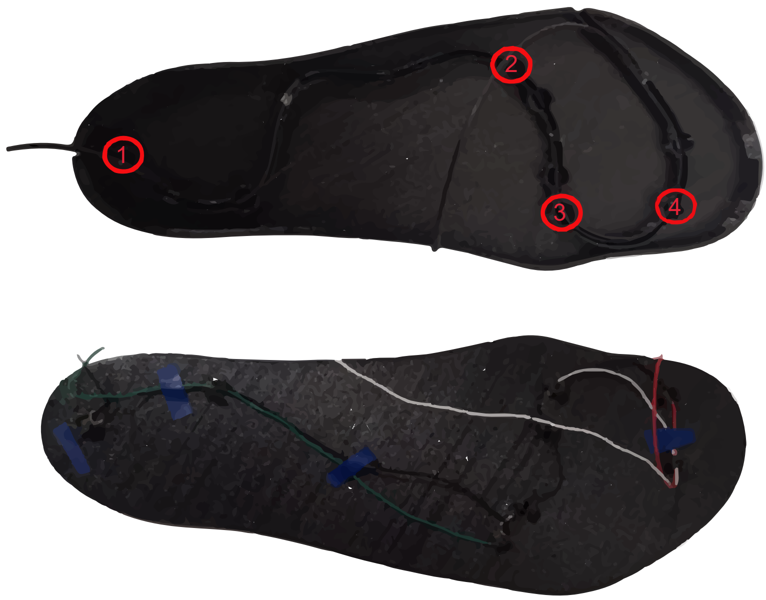


Figura 1: Instrumented insole with the optical component built-in. In the upper part, it is the insole base with the optical fiber and in the lower part it is the insole cap with the LEDs. The circled points are the sensitive zones positions.

minutes; Once this is done, the next step is to pour the silicon into the insole empty spaces; after the cure time that is approximately four hours, the silicone became solid and, the insole cap was attached to the base insole by sewing the insole base to the insole cap with hemp. The sewing was done in strategical locations of the insole contour to avoid limiting the flexible adaptation of the insole to movements during gait.

3.2. Acquisition and Transmission Modules

These modules have different components, they are the ones that controls devices that make part of the insole and interconnects the instrumented insole with the processing unit. The acquisition system is formed by a development platform FRDM KL25Z, an optical fiber coupling board, an array of LEDs and, the POF fiber that as was described in previous subsections were incorporated to the insole.

The FRDM KL25Z is programmed with the LEDs' switching-on routine to generate the pressure signal, thereby, this board is connected to the optical fiber coupling board. FRDM KL25Z also makes the ADC reads of the two photodetectors that are in the fiber coupling board. The acquisition system also includes an electrical bakelite circuit board which is the optical fiber coupling board. This is made up of a transistor arrangement (2N3906 PNP) which function is to connect the logic outputs of the development board to the LEDs; and two phototransistors (IF D92) to measure the LEDs' optical power. Phototransistor 1 is near lateral section 1 and, phototransistor 2 is near lateral section 4.

The transmission module involves the FRDM KL25Z, a HC-05 module and a processing unity to store the data. The ADC reads of the photodetectors read by the FRDM KL25Z are sent by this same board to a laptop that receives, stores the data in *.txt* files generated by the processing unity Python, and generates graphs the data. The data

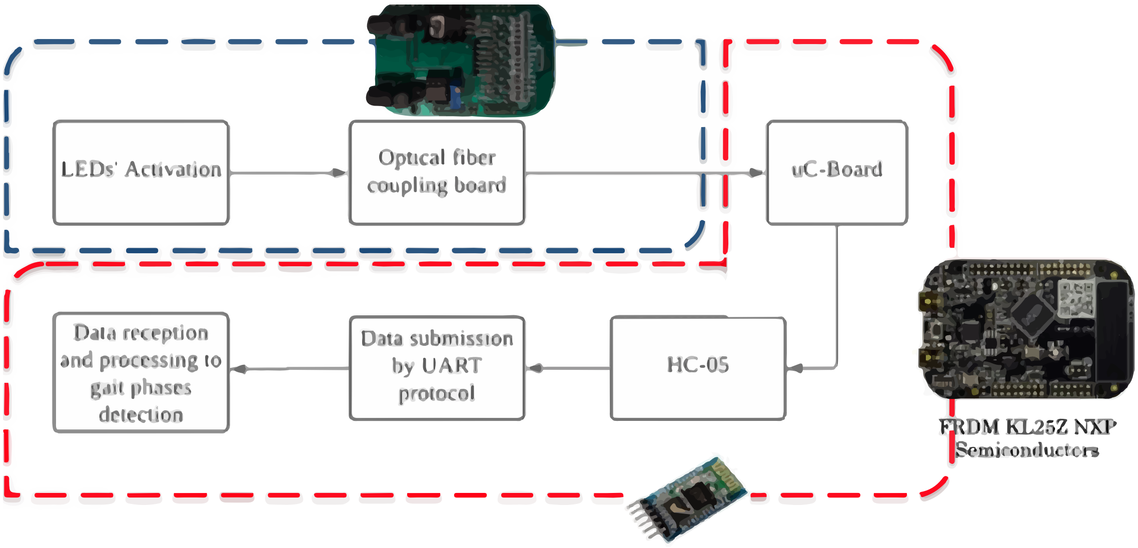


Figura 2: Acquisition and transmission module with its components. The part in the blue shape is the acquisition component and the one in the red shape is the transmission component.

sending is done through the HC-05 module (Bluetooth). From the FRDM KL25Z, the data are sent with a baud rate of 115200. Since the HC-05 has a resolution of 8 bits, the analog read of the light intensity is first fragmented and then sent. Therefore each datum has to be rebuilt in python. The acquisition and transmission system is in the diagram shown in Fig. 2.

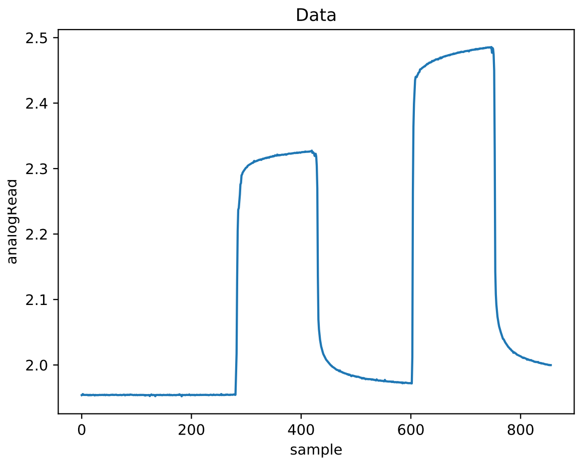
3.3. Lateral Sections Characterization Setup

As this is a sensor to measure plantar pressure, it was tested with a series of calibrated weights of $1kg$, $2kg$, $5kg$, and $10kg$, following a protocol that contemplates $10s$ of sensor's offset, $10s$ of weight positioning and $5s$ of sensor's recovery (exchange time between weights). Some tags were made on the outer surface of the insole cap to execute this protocol providing repeatability; they mark the LED and sensitive zone position to place the loads right in the center of the zones. The insole was placed on a flat surface and, the described protocol was applied to each sensitive zone and, the phototransistors response was characterized independently (in different instants of time).

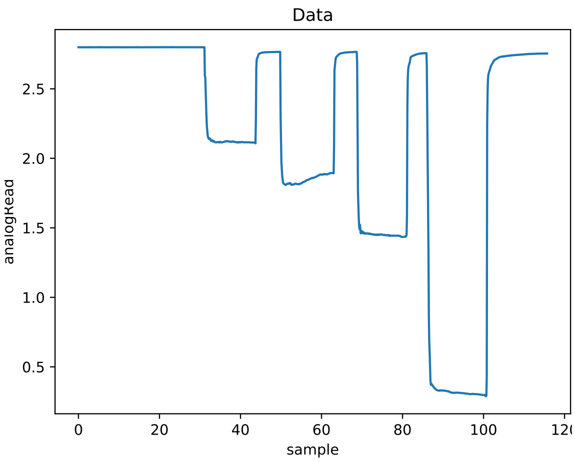
The LEDs are lighted up whit the FRDM KL25Z board with a $30Hz$ frequency (according to the LEDs tuning-on routine). The signal acquisition during the characterization process was performed with the FRDM KL25Z and with a USB cable serial connection openSDA to the PC. The stored data in .txt files were plotted to verify the sensor performance when the intensity was measure with phototransistors 1 and 2. The silicone arranged around the fiber path afford a better and even weight distribution, allowing to reduce the influence of the human factor due to the fatigue and the force addition in the characterization process.

4. RESULTS

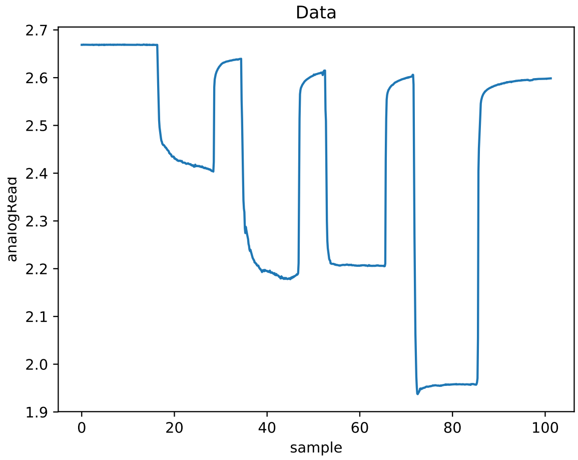
This section presents the results of the sensitive zones characterization and the linear behavior of each one of this zones. There are two different results of each sensitive zone, the outcome of photodetector 1 and the outcome of the photodetector 2. Fig. 3 and Fig. 4 present the response to the load of the lateral sections measured with phototransistors 1 and 2 respectively. The expected behavior of the sensitive zones is that the voltage value decreases proportionally to the load increase. This behavior is present in sensitive zones 2 (Fig. 3b and 4a), 3 (Fig. 3c and 4b) and 4 (3d and 4c). However, As can be appreciated in Fig. 3a, the sensitive zone 1 does not have this behavior because the voltage is increasing instead of decreasing with load positioning.



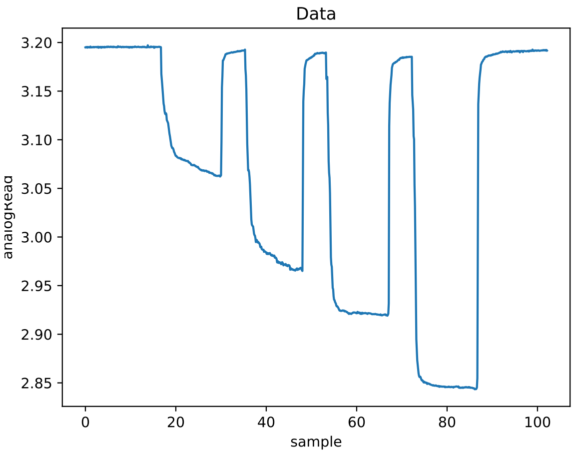
(a) Sensitive zone number 1



(b) Sensitive zone number 2



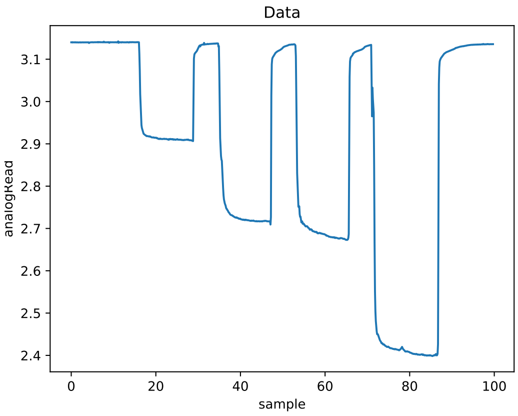
(c) Sensitive zone number 3



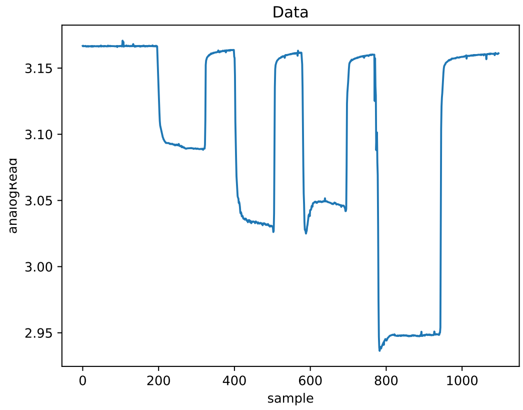
(d) Sensitive zone number 4

Figura 3: Characterization of the sensitive zones using the phototransistor 1.

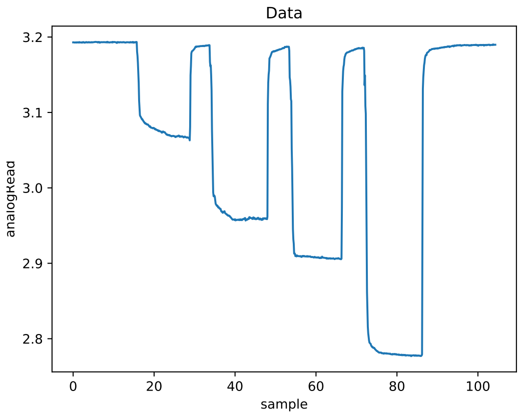
Based on the results shown in Fig. 3 and Fig. 4 graphs in Fig. 5 and Fig. 6 were generated; in these graphs, the sensitive zones' response to load can be better appreciated in terms of linearity. Fig. 6b shows the linearity curve with the least $R^2 = 0{,}61$; this behavior corresponds to the zone 3 with photodetector 1. The maximum R^2 of the measurements made with phototransistor 1 belongs to the lateral section 1 and it is $R^2 = 0{,}95$, this is the nearest point to that photodetector. However it is necessary to consider that this estimation was done with just two points. In the case of phototransistor 2, the maximum and minimum R^2 are $R^2 = 0{,}8578$ and $R^2 = 0{,}8045$ respectively. The highest R^2 belongs to the sensitive zone which is the nearest to phototransistor 2.



(a) Sensitive zone number 2



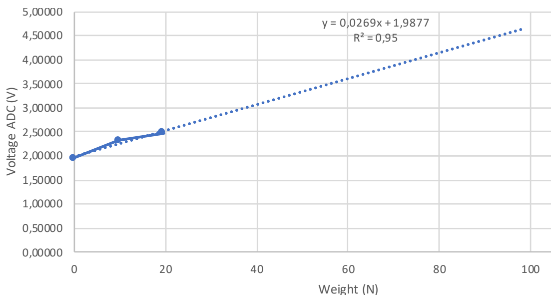
(b) Sensitive zone number 3



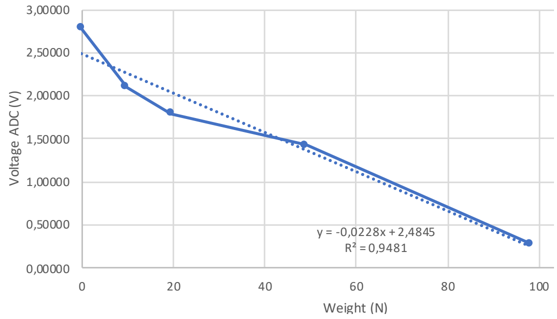
(c) Sensitive zone number 4

Figura 4: Characterization of the sensitive zones using the phototransistor 2.

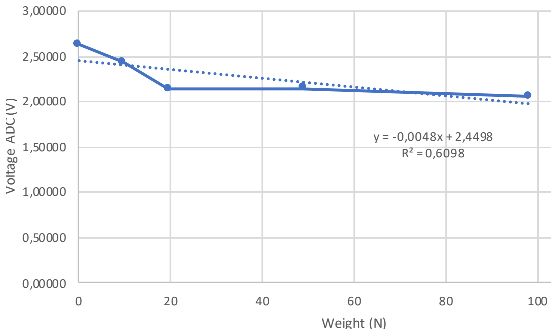
The behavior shown in these curves is influenced by the distance from the sensitive zone to the photodetectors and the loads positioning; thus the voltage variation is less when the lateral section is further and it is less when the weigh distribution area is bigger.



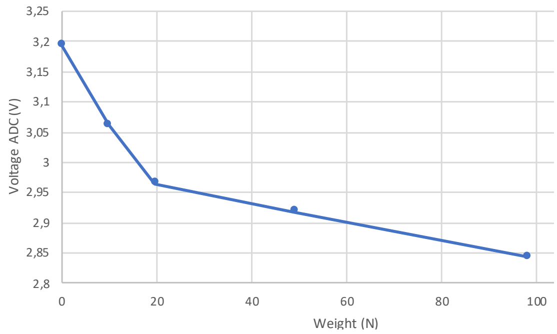
(a) Sensitive zone number 1 response



(b) Sensitive zone number 2 response

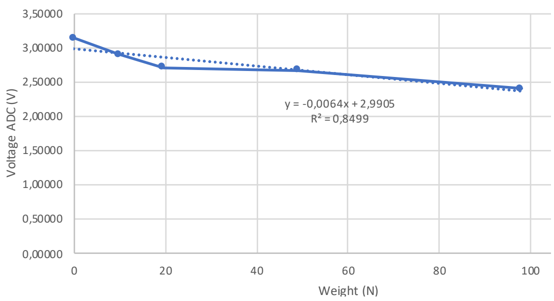


(c) Sensitive zone number 3 response

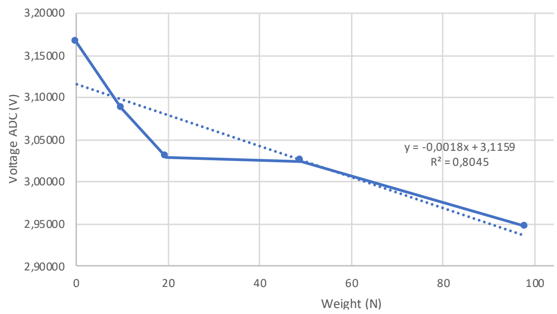


(d) Sensitive zone number 4 response

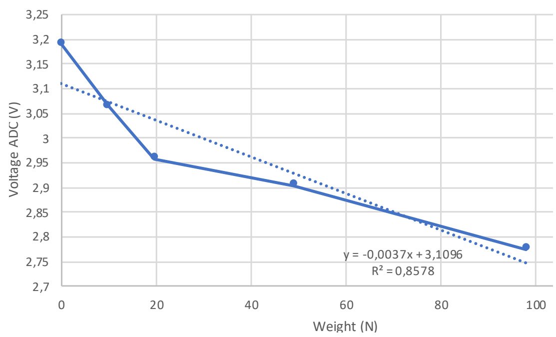
Figura 5: Sensitive zones response to load reading phototransistor 1.



(a) Sensitive zone number 2 response



(b) Sensitive zone number 3 response



(c) Sensitive zone number 4 response

Figura 6: Sensitive zones response to load reading phototransistor 2.

5. DISCUSSION

Many factors can affect the answer of the sensitive zones to loads. One of this is the fact that the sensitive zones are handmade, therefore, there are differences in the shape and depth of these zones due to the human factor and these differences affect the sensitivity of the fiber. Another observed factor is the insole material deformation: some parts of the material that come of can cause interference between the light emitted by the LED and the sensitive zone. On the other hand, the attachment of the insole cap to the base can allow the ambient light to cause interference.

With the applied characterization protocol the sensor's response to loads and their linearity was determined. Three of the four lateral sections showed the expected behavior, which is a lineal response to loads with a high value of R^2 and the light intensity decreasing with the weight increment. In the case of phototransistor 1, the linear fitting curve did not reach R^2 greater than $0{,}95$, and in the case of phototransistor 2, it does not exceed $0{,}85$. In literature POF sensors that reach values no lower than $R^2 = 0{,}990$ are exposed. However, the authors use photodetectors that are near by the sensitive zones to characterize the sensors. In the developed instrumented insole, the R^2 highest values where found in the closest sensitive zones to the photodetectors.

In Fig. 3a and Fig. 4b a behavior in which the voltage increases with weight increase can be appreciated. this behavior can occur due to problems related to the insole setup; in those cases, the fiber can not be correctly aligned with the light source, or something can be hindering the passage of the LED's light to the sensitive zone. These facts remark the necessity of a compact and robust system that presents a proper attachment of the light sources and the lateral sections of the fiber to guarantee repeatability and accurate measurements. Additionally, there is a need to ensure homogeneity in the fiber modification to generate the sensitive zones.

6. FUTURE WORKS

Through this work an instrumented insole to measure plantar pressures and its acquisition and transmission modules were developed, what was achieved was to generate the whole structure of the instrumented insole, the measurement setup an a characterization of the developed POF sensor. From this, also an evaluation of the insole reliability regarding the technical, mechanical, and assessment requirements is necessary.

For being able to measure the aforementioned, a gait events detection algorithm has to be developed, validated and assessed. After this, the insole assessment should include the comparison with an existent technology (an instrumented insole developed with sensors that have a different operation principle) implemented for gait events assessment. An option is to test the insole with healthy individuals and comparing the results with an instrumented insole with four FSR sensors in the same position as the fiber's sensitive zones. Finally the insole could be tested with individuals with a pathology that affects the gait pattern.

7. CONCLUSIONS

In this work, an instrumented insole with a pressure sensor based on a polymeric optical fiber, and measurement of power intensity by the multiplexing technique was presented. This instrumented insole was complemented with an acquisition an transmission module to acquire and store pressure measurements and to do a characterization of the different sensitive zones elaborated in the fiber. The developed instrumented insole is composed of a cap and a base where the measurement system (the optical fiber and the array of LEDs) is nested. This instrumented insole is connected the acquisition and transmission modules that includes two different boards, the FRDM KL25Z and the optical fiber coupling board, and the HC-05 module.

The functioning of the instrumented insole was programmed in the microcontroller of the FRDM KL25Z. The microcontroller executes a LEDs' turn-on routine according to a multiplexing technique at a frequency of 30 Hz. A code for storing the data was developed in the chosen processing unity that was Python. This code received from the FRDM KL25Z an through the HC-05 module a fragmented datum of 8 bits; then, datum was reconstructed and stored for being plotted and processed. Finally a characterizartion protocol was designed and implemented to calibrate the four sensitive zones and determine their response to load and their linear correlation of this response.

Referencias

[1] J. Cardoso, C. R. J. Alves, D. Baptista, C. D. Sartor, A. H. Nascimento, W. Marques, E. Z. Martinez, C. Isabel, N. Sacco, and A. C. Mattiello-sverzut, "Gait & Posture Dynamic plantar pressure patterns in children and adolescents with Charcot-Marie-Tooth disease," *Gait & Posture*, vol. 86, no. November 2019, pp. 112–119, 2021.

[2] S. Tashiro, N. Gotou, Y. Oku, T. Sugano, T. Nakamura, H. Suzuki, N. Otomo, S. Yamada, T. Tsuji, Y. Asato, and N. Ishii, "Relationship between plantar pressure and sensory disturbance in patients with hansen's disease—preliminary research and review of the literature," *Sensors (Switzerland)*, vol. 20, no. 23, pp. 1–18, 2020.

[3] A. P. Konings-Pijnappels, M. Tenten-Diepenmaat, R. Dahmen, S. K. Verberne, J. Dekker, J. W. Twisk, L. D. Roorda, and M. van der Leeden, "Forefoot pathology in relation to plantar pressure distribution in patients with rheumatoid arthritis: A cross-sectional study in the Amsterdam Foot cohort," *Gait and Posture*, vol. 68, no. December 2018, pp. 317–322, 2019.

[4] A. Yildiz, E. Yildirim, O. Ozturk, I. Demirbuken, M. Ozturk, and M. G. Polat, "P 142 – Plantar pressure distributions in patients with chronic obstructive pulmonary disease," *Gait and Posture*, vol. 65, no. 2018, pp. 473–474, 2018.

[5] J. Hu, H. Cao, Y. Zhang, and Y. Zheng, "Wearable Plantar Pressure Detecting System Based on FSR," *Proceedings of 2018 2nd IEEE Advanced Information Management, Communicates, Electronic and Automation Control Conference, IMCEC 2018*, no. Imcec, pp. 1687–1691, 2018.

[6] J. A. Ramirez-Bautista, A. Hernández-Zavala, S. L. Chaparro-Cárdenas, and J. A. Huerta-Ruelas, "Review on plantar data analysis for disease diagnosis," *Biocybernetics and Biomedical Engineering*, vol. 38, no. 2, pp. 342–361, 2018.

[7] F. Deligianni, C. Wong, B. Lo, and G.-z. Yang, "A fusion framework to estimate plantar ground force distributions and ankle dynamics," no. January, pp. 255–263, 2020.

[8] R. D. Gurchiek, C. P. Garabed, and R. S. Mcginnis, "Gait event detection using a thigh-worn accelerometer," *Gait & Posture*, vol. 80, no. June, pp. 214–216, 2020.

[9] E. Shahabpoor and A. Pavic, *Measurement of walking ground reactions in real-life environments: A systematic review of techniques and technologies*, vol. 17. 2017.

[10] R. Mann, L. Malisoux, A. Urhausen, K. Meijer, and D. Theisen, "Plantar pressure measurements and running-related injury: A systematic review of methods and possible associations," *Gait and Posture*, vol. 47, pp. 1–9, 2016.

[11] I. Klöpfer-Krämer, A. Brand, H. Wackerle, J. Müßig, I. Kröger, and P. Augat, "Gait analysis – Available platforms for outcome assessment," *Injury*, vol. 51, no. xxxx, pp. S90–S96, 2020.

[12] C. S. Morère, A. Sura, A. R. Pérez-tabernero, E. Vihriälä, and T. Myllylä, "MEMS Technology Sensors as a More Advantageous Technique for Measuring Foot Plantar Pressure and Balance in Humans," vol. 2016, 2016.

[13] E. Martini, F. Tommaso, F. Dell Angello, Z. Ivanic, M. Munih, N. Vitiello, and S. Crea, "Pressure-Sensitive Insole for Real-Time Gait-Related Applications," *Sensors (Switzerland)*, 2020.

[14] A. G. Leal-junior, M. F. Domingues, R. Min, A. Frizera-neto, P. Andr, and P. Antunes, "Fiber Bragg Based Sensors for Foot Plantar Pressure Analysis," pp. 3–25.

[15] J. A. Ramirez-Bautista, J. A. Huerta-Ruelas, S. L. Chaparro-Cárdenas, and A. Hernández-Zavala, "A Review in Detection and Monitoring Gait Disorders Using In-Shoe Plantar Measurement Systems," *IEEE Reviews in Biomedical Engineering*, vol. 10, pp. 299–309, 2017.

[16] J. Figueiredo, C. Ferreira, L. Costa, J. Sepulveda, L. P. Reis, J. C. Moreno, and C. P. Santos, "Instrumented insole system for ambulatory and robotic walking assistance: First advances," *2017 IEEE International Conference on Autonomous Robot Systems and Competitions, ICARSC 2017*, pp. 116–121, 2017.

[17] S. Vigneshwaran and G. Murali, "Foot plantar pressure measurement system for static and dynamic condition," *IOP Conference Series: Materials Science and Engineering*, vol. 993, no. 1, 2020.

[18] Z. You, A. Zahid, H. Heidari, M. Imran, and Q. H. Abbasi, "A Compact Wearable System for Detection of Plantar Pressure for Diabetic Foot Prevention," *Asia Pacific Conference on Postgraduate Research in Microelectronics and Electronics*, vol. 2018-Octob, pp. 64–67, 2018.

[19] A. Jor, S. Das, A. S. Bappy, and A. Rahman, "Foot Plantar Pressure Measurement Using Low Cost Force Sensitive Resistor (FSR): Feasibility Study," *Journal of Scientific Research*, vol. 11, no. 3, pp. 311–319, 2019.

[20] G. Rescio, A. Leone, L. Francioso, and P. Siciliano, "Sensorized Insole for Diabetic Foot Monitoring," *Proceedings*, vol. 2, no. 13, p. 860, 2018.

[21] A. S. Morris and R. Langari, "Sensor Technologies," in *Measurement and Instrumentation*, pp. 375–405, Elsevier, jan 2016.

[22] C. Gerlach, D. Krumm, M. Illing, J. Lange, O. Kanoun, S. Odenwald, and A. Hubler, "Printed MWCNT-PDMS-Composite Pressure Sensor System for Plantar Pressure Monitoring in Ulcer Prevention," *IEEE Sensors Journal*, vol. 15, no. 7, pp. 3647–3656, 2015.

[23] W. Bautista-aguiar, D. Florez-quintero, D. Narvaez-martinez, and S. H. C.-o. B, "Instrumented Insole for Plantar Pressure," pp. 252–259, 2018.

[24] C. Lou, S. Wang, T. Liang, C. Pang, L. Huang, M. Run, and X. Liu, "A graphene-based flexible pressure sensor with applications to plantar pressure measurement and gait analysis," *Materials*, vol. 10, no. 9, 2017.

[25] Q. Zhang, Y. Lu, Y. Xia, X. Wu, T. Vernon, and X. Dong, "A low-cost and highly integrated sensing insole for plantar pressure measurement," *Sensing and Bio-Sensing Research*, vol. 26, no. September, 2019.

[26] L. Motha, J. Kim, and W. S. Kim, "Instrumented rubber insole for plantar pressure sensing," *Organic Electronics*, vol. 23, pp. 82–86, 2015.

[27] S. Crea, M. Donati, S. M. M. De Rossi, C. Maria Oddo, and N. Vitiello, "A wireless flexible sensorized insole for gait analysis," *Sensors (Switzerland)*, vol. 14, no. 1, pp. 1073–1093, 2014.

[28] S. W. Park, P. S. Das, and J. Y. Park, "Development of wearable and flexible insole type capacitive pressure sensor for continuous gait signal analysis," *Organic Electronics*, vol. 53, no. November 2017, pp. 213–220, 2018.

[29] P. Aqueveque, E. Germany, R. Osorio, and F. Pastene, "Gait segmentation method using a plantar pressure measurement system with custom-made capacitive sensors," *Sensors (Switzerland)*, vol. 20, no. 3, 2020.

[30] H. Kadhum Hisham, "Optical Fiber Sensing Technology: Basics, Classifications and Applications," *American Journal of Remote Sensing*, vol. 6, no. 1, p. 1, 2018.

[31] E. Herriko, *Desarrollo de un sensor de fibra óptica para la medida del tip clearance y tip timing en motores aeronáuticos*. PhD thesis, 2017.

[32] M. F. Domingues, N. Alberto, C. Leitão, C. Tavares, E. R. D. Lima, A. Radwan, V. Sucasas, J. Rodriguez, P. André, and P. Antunes, "Insole optical fiber sensor architecture for remote gait analysis - an eHealth Solution," vol. 4662, no. c, 2017.

[33] A. G. Leal-junior, A. Frizera, L. M. Avellar, and C. Marques, "Polymer optical fiber for in - shoe monitoring of ground reaction forces during the gait," vol. 1748, no. c, 2018.

[34] A. G. Leal-Junior, A. Frizera, L. Vargas-Valencia, W. M. Dos Santos, A. P. Bo, A. A. Siqueira, and M. J. Pontes, "Polymer Optical Fiber Sensors in Wearable Devices: Toward Novel Instrumentation Approaches for Gait Assistance Devices," *IEEE Sensors Journal*, vol. 18, no. 17, pp. 7085–7092, 2018.

[35] Y. Haseda, J. Bonefacino, H. Y. Tam, S. Chino, S. Koyama, and H. Ishizawa, "Measurement of pulse wave signals and blood pressure by a plastic optical fiber FBG sensor," *Sensors (Switzerland)*, vol. 19, no. 23, 2019.

[36] A. Leal-junior, L. Avellar, J. Jaimes, D. Camilo, W. Santos, A. A. G. Siqueira, M. Jos, and C. Marques, "Polymer Optical Fiber-Based Integrated Instrumentation in a Robot-Assisted Rehabilitation Smart Environment : A Proof of Concept," pp. 1–16, 2016.

[37] D. Sartiano and S. Sales, "Low cost plastic optical fiber pressure sensor embedded in mattress for vital signal monitoring," *Sensors (Switzerland)*, vol. 17, no. 12, 2017.

[38] A. G. Leal-Junior, C. R. Díaz, C. Marques, M. J. Pontes, and A. Frizera, "3D-printed POF insole: Development and applications of a low-cost, highly customizable device for plantar pressure and ground reaction forces monitoring," *Optics and Laser Technology*, vol. 116, no. January, pp. 256–264, 2019.

[39] D. Vilarinho, A. Theodosiou, C. Leitão, A. G. Leal-Junior, M. de Fátima Domingues, K. Kalli, P. André, P. Antunes, and C. Marques, "POFBG-embedded cork insole for plantar pressure monitoring," *Sensors (Switzerland)*, vol. 17, no. 12, 2017.

[40] A. G. Leal-Junior, C. R. Díaz, C. Marques, M. J. Pontes, and A. Frizera, "Multiplexing technique for quasi-distributed sensors arrays in polymer optical fiber intensity variation-based sensors," *Optics and Laser Technology*, vol. 111, no. August 2018, pp. 81–88, 2019.

[41] A. G. Leal-Junior, A. Frizera, and M. José Pontes, "Sensitive zone parameters and curvature radius evaluation for polymer optical fiber curvature sensors," *Optics and Laser Technology*, vol. 100, pp. 272–281, 2018.