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Soft Sensory Socks Measure Contact Forces During Locomotion

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Abstract— **Gait monitoring systems play a crucial role in guiding rehabilitation, improving athletic performance, preventing injuries, and ensuring overall mobility and quality of life. Most of these systems are restricted to specialized setups and require calibrations. We propose a soft sensor design using silicone tubes mounted on the bottom side of socks to measure ground reaction force (GRF) during locomotion. Three sensors were mounted on the socks' anterior, lateral, and posterior areas on the left and right feet. The sensory socks enable the analysis of gait while walking and running at different speeds while wearing shoes and also barefoot. The proposed soft sensory socks open the door to analyzing locomotion in different environments and also barefoot. Preliminary gait analyses confirm the different gait parameters observed in the biomechanics of barefoot locomotion.**

I. Introduction

Gait monitoring systems are used in healthcare, sports, and rehabilitation to analyze the way people walk and run, providing valuable data about an individual's movement patterns, which can be critical for diagnosing health conditions, improving physical performance, and preventing injuries [1], [2]. Gait monitoring systems using ground reaction force (GRF) offer the ability to track different gait phases, to detect gait abnormalities, and can further be used in exoskeleton systems as a form of external input [3]. In recent years a number of soft wearable sensor systems have been proposed. Compared to professional lab equipment, these systems offer advantages in terms of costs and applicability [1]. Given the use of such systems in rehabilitation, the necessity of material endurance and reliability in terms of repeated usage and signal quality plays an ever-so-important role. Wearable soft GRF sensors can be broadly classified into two categories: electric sensors and pneumatics-based sensors. Electric sensors measure GRFs by the direct force effect on electric components in the circuit. Examples include force sensing resistors (FSRs) and capacitive sensors, which offer a robust method of force measurement in different conditions [4], [5]. While traditional electronics, such as FSRs, are reliable in measuring normal forces, they quickly reach their practical limits when used in a sole for measuring GRFs for gait analysis. By having rigid components, these are prone to breakage over time as soles bend during the gait cycle. A number of pressure-based systems have also been proposed in recent years. For example, an air pressure-based force sensor sole has been developed by Kong and Tomizuka [6]. The design uses silicone tubes with an outer diameter of 4mm mounted on a custom shoe, thus allowing the detection

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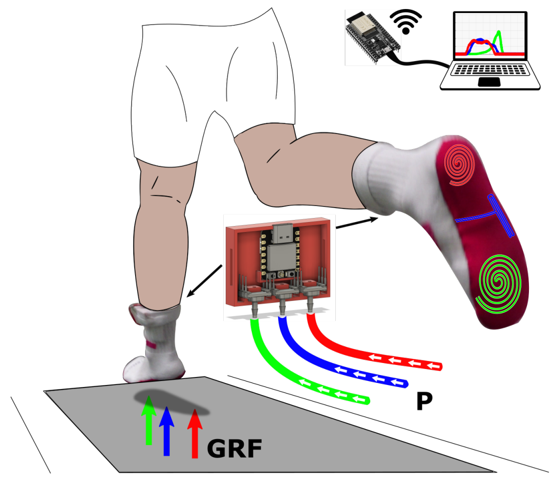

Fig. 1: Overview of the soft sensory socks while running barefoot on a treadmill during the floating phase. The ground reaction forces at each foot are measured by three sensors and translated into the air pressure of the silicone tubes.

of the gait cycle during walking. A similar method has been proposed in [7]. However, the proposed system can not measure GRFs during locomotion in barefoot walking. Barefoot walking shows increasing advantages for the biomechanics of the feet and legs compared to walking with shoes, such as reduced loads on lower extremity joints [8]. Nevertheless, wearing shoes is an integral part of current daily life. To better understand the effects of the common shoe on joint health across the lifespan, more research on barefoot walking in older age populations (50+ years) is needed. However, currently there is a lack of such research [9]. One reason may be due to a potential lack of convenient solutions to precisely measure GRFs during barefoot walking. Therefore, we propose a new sensory sock design that bypasses this issue. It uses 1.3mm outer diameter silicone tubes directly attached to a sock using a 1mm thick silicone layer. Similar silicone tubes have been used in [10] as variable resistors to study hand movements. By covering the open side with cloth, the sensory sock can comfortably fit into more shoe types and adhere its shape to fit to existing soles due to the flexibility and thinness of the sensor layer. Thus, the design of the sock is independent of specific shoes, and the sensory sock can even be used to collect information on barefoot walking.

The paper is organized as follows. Section II presents the design and benchmarking of the soft force sensor. Section III illustrates the design of the soft sensory socks. A preliminary measurement for gait estimation is presented in Section IV.

Section V presents the conclusion of the paper.

II. Soft Force Sensor

A. Design and Working Principle

1) Design: A molding was designed such that silicone tubes with an inner tube diameter of 0.5mm and a wall thickness of 0.4mm can be placed in a radial pattern (see Fig.5) as such a pattern is known to be efficient in measuring normal forces [6]. With the tube placed inside the molding, silicone is poured to completely fill the molding. To hasten the curing process and improve the silicone's ultimate tensile strength, the silicone is left to cure at a temperature of 48°C for an hour. Once cured and the molding removed, the silicone tubes are protected by a layer of silicone on one side. To fully protect the tube from tearing, the same procedure is repeated on the still exposed side to fully enclose the tubes. The silicone used is the Dragon Skin 10 SLOW from Smooth-On. Ideally, the tube has silicone layers of equivalent thickness on both sides. Furthermore, it protrudes out of the mold, with one end long and one short. The long end is 32cm long and attached to the air pressure sensor. The short end is tied into a knot to prevent the airflow outside the tube. The pressure sensor used is the ABPDANT005PGAA5 from Honeywell. With a measurement range of 0-5psi and precision of 0.25% makes it a reliable sensor. Furthermore, it is also a human centered design choice as its small form factor makes it convenient and comfortable to wear. The soft sensor made to collect the measurements for the sections below had a total tube length of 124cm with a long end of 32cm and short end of 14cm. The actual thickness of the soft sensor was 3.4mm.

2) Working Principle: With tubes placed within a silicone patch, it is assumed that these cannot change shape to maintain equal air pressure when a force is exercised upon the patch to due to the relative rigidity of the silicone patch. Hence, when a normal force is applied on the silicone patch, the tubes are compressed in volume. As there is no open end for the air to escape, this leads to an increase in pressure within the tubes, which can be measured by the pressure sensor attached to one end.

B. Sensor Characteristics

1) Test Bench: To verify the performance of the soft sensor, a test bench was made using the PSD-S1 high precision load cell (50kg capacity) from Pushton as the reference force sensor and the DSNU-25-100-P-A-S11 ISO cylinder from Festo as actuator and the MPYE-5-1/8-HF-010-B proportional directional control valve from Festo to control said actuator. The force of the actuator can be controlled using PWM signals, which control the airflow into the actuator. As this is a pneumatic actuator, the cylinder was attached to an air compressor that had an output of 6 bars(87 PSI).

To simultaneously record data from the load cell and from the soft sensor, a wooden plate was attached to the load cell and the soft sensor was placed on top of this plate. The actuator had a cylindrical end effector with a contact diameter of 4 cm and was placed perpendicularly such that it would press down on the sensor and load cell together.

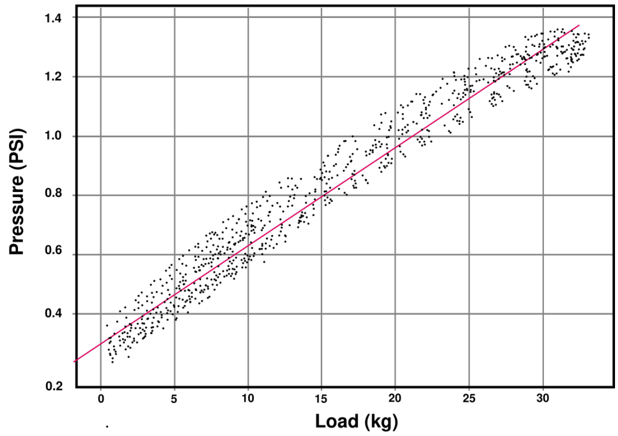

Fig. 2: Load cell readings versus the soft sensor pressure readings. The plot suggests a linear correlation between the applied foce and soft sensor readings.

2) Force Correlation and Hysteresis: To use the soft sensors as a force-measuring sensor, the pressure readings of the sensor must ideally be linear to the applied normal force. To verify this, the actuator was oscillated between 0 kg and 33 kg with a step resolution of 29 steps between 0 kg and 33 kg. Each step was held for 1.5 s to allow the sensor enough time to measure steady state values at each exerted force step.

the actuator was gradually oscillated between two set force values with a period of 87s such that quasi-static force readings could be measured to prevent any dynamic effects from occurring.

Figure 2 shows an approximation of the linear correlation between the soft sensor pressure and load cell readings, a linear regression giving the following:

$$F(p) = 0.032\frac{kg}{psi}p + 0.221kg \tag{1}$$

where p is the measured air pressure from the soft sensor in PSI and F is the force measured by the load cell in kg.

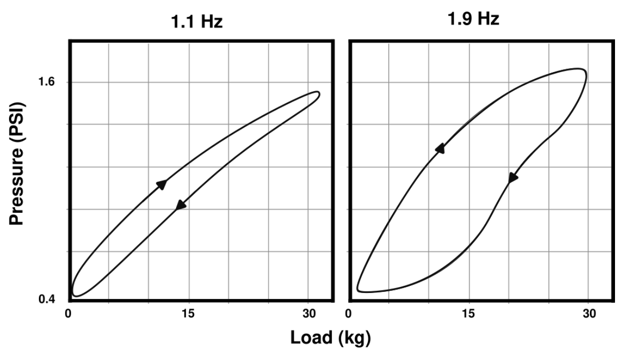

Fig. 3: The hysteresis of the soft sensor at an actuator oscillation frequency of 1.1 Hz and 1.9 Hz respectively.

In addition to the force correlation, hysteresis at higher frequencies was measured to verify the soft sensor's performance at higher frequency stimulations such as when

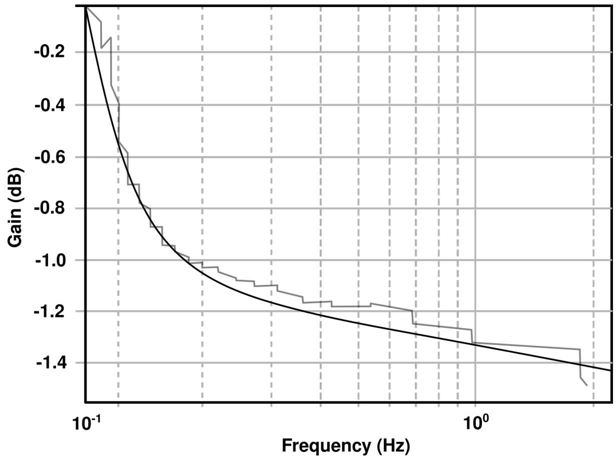

Fig. 4: Frequency response of the proposed soft force sensor.

running. With common human stride frequencies ranging from 0.7 Hz to 1.5 Hz [11], where 0.7 Hz tends to indicate slow walking and 1.5Hz running, actuator frequencies of roughly 1 Hz and 2 Hz were tested on the sensor to verify its viability for common walking speeds and test its stability for fast gait cycles. Figure 3 shows a narrower hysteresis at 1Hz compared to 2Hz, indicating that the soft sensor is viable for measuring gait characteristics when walking.

3) Frequency Response: With initial hysteresis data looking promising for gait analysis, the frequency response for a wider range of stimulation frequencies from 0.1 Hz to 5 Hz was measured. The gain for the response was calculated by comparing the reduction in amplitude of the reference load cell and soft sensor over time.

The frequency response is shown in Fig.4 and an exponential decay in signal amplitude is visible with increasing frequency. The actuator used to test the sensor could not produce maximum oscillations at frequencies higher than 2 Hz. This may have contributed to the decrease in gain at higher frequencies. Nevertheless, with a decrease not exceeding -1.3 dB, the soft sensor is certainly viable to use within the common stride frequencies of human gait, even for running.

III. Sensory Socks

A. Manufacturing

Using the soft sensor principle explained in Sec. II, a pair of sensory socks were made, each containing three such sensor patches. One to measure the contact forces on the anterior of the foot, one for the lateral, and one channel for the posterior of the foot. To place the silicone tubes in their desired location and shape, a molding out of PLA was 3D printed in the shape of a EUR size 42 sole with grooves placed accordingly. The anterior and posterior silicone tubes are placed in spiral structures, while the lateral tube is placed in a longitudinal structure. With the silicone tubes placed into the groove, roughly 50 g of silicone is poured over them for each sock. A 3D printed plate, in the same shape as the silicone molding is placed inside the sock and the sock is

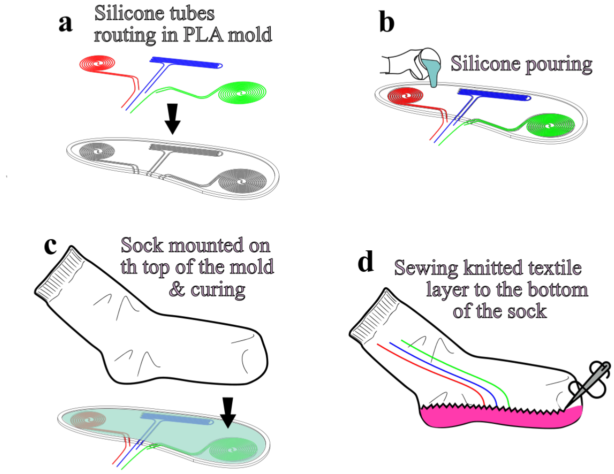

Fig. 5: Soft sensory socks manufacturing steps. With the silicone tubes placed inside the molding, silicone is poured on top and then the sock is placed on top and left to cure. Then, a knitted fabric layer is mounted on the bottom of the sock to protect the three sensor patches from damage during barefoot locomotion.

then placed on top of the silicone. Different to the sensor patch in Sec. II, the second layer of silicone is replaced by a cloth. This is done in favor of an even thinner sensory sock. Additionally, the smooth surface of the cloth makes the sock versatile to wear with various shoe types. Figure 5 shows the preparation steps of the sensory sock. Overall, around 50g of silicone is needed to make a silicone layer for one sock. The sensors have different silicone tube lengths in function to the contact area they cover and the routing path to the air pressure sensors. The tube length of the anterior pressure sensor is 162cm, and the diameter of the formed spiral is 6.1cm. The tube length of the lateral pressure sensor is 99cm, and the dimensions of the formed rectangle are 8.9cm by 1.4cm. The tube length of the posterior pressure sensor is 103cm, and the diameter of the formed spiral is 4.1cm.

B. Data Acquisition

The silicone tubes of each sensory sock were attached from one side to three air-pressure sensors (Honeywell ABPDANT005PGAA5) whereas the other sides were tied into knots to prevent air leakage. The air-pressure sensors were placed into a box and then clipped onto the top of the sock, as can be seen in Fig.1. The sensor output is analog and can measure within the range from 0 to 5 PSI. The analog signals of the air pressure sensors were measured by the "XIAO ESP32C3" microcontroller, which has a WiFi module. Each sock has a dedicated microcontroller sending readings into an ESP32 server over WiFi, with a sending rate of 100 Hz for each sensor. A similar WiFi setup has been adapted in [12]. The ESP32 server then sends the data into a PC by serial port. "CoolTerm", a freeware serial port terminal software, was used to store the received data in text files.

IV. Gait Measurement

A. Common Gait Characteristics

In general, there are a variety of biomechanical features that can be used to characterize gait in a human, with certain features being specific to barefoot locomotion. As described by Houglum [13] the healthy human gait cycle is grouped into phases. Figure 6 provides an overview of these phases for walking and running. The walking gait cycle consists of a stance phase preceded by a swing phase. The stance phase is described by the contact of the foot of interest to the ground while the other leg is in swing phase. Key features of the stance phase are the heel-strike, the mid-stance, and the toe-off. After the toe-off, the foot of interest then enters the swing phase, where the leg swings forward while the other is now in stance phase. The running gait cycle is accompanied by two additional float phases. In the float phase, neither foot is in contact with the ground. When running, the stance phase is followed by a floating phase where the foot of interest is kicked back and the other foot is in front. This is followed by a swing phase where it swings forward while the other foot is in stance phase, and enters another float phase kicking in front before finally contacting the ground for stance phase.

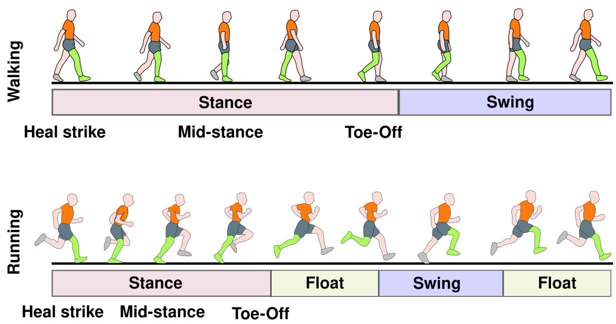

Fig. 6: Visualization of different phases and key time points of the gait cycle during walking and running.

It is also well-established that there are significant differences between barefoot locomotion and locomotion with shoes. Kyle et al. [14] reports that the ground contact time of the foot is shorter in barefoot locomotion compared to locomotion with shoes and a shorter stride length is also observed. Furthermore, in barefoot locomotion smaller maximum vertical ground reaction forces are observed. A diverse amount of information can be obtained from the analysis of such features. Shakoor et al. [8], for example, observed a stronger stress on low extremity joints in walking with shoes compared to walking barefoot. Another surprising observation is the significant difference in the symmetry of the locomotion. As reported in multiple works [15], [16], barefoot locomotion is accompanied by a larger asymmetry between the feet than in locomotion with shoes. Hence, the closer analysis of gait characteristics with and without shoes can, for example, provide key insights into the efficacy of shoes in long-term joint health. This type of information may even prove beneficial in the decision-making of therapy.

B. Gait monitoring with soft sensory socks

In order to verify whether the proposed soft sensory socks can measure the gait characteristics described above, a single male participant of 72kg and 172cm, with a EU shoe size of 42, was asked to wear the sensory socks and walk on a treadmill at different speeds. It started at a speed of 2 km/h and manually increased by 2 km/h every 40 seconds until it reached 12 km/h. The sensor data for different locomotion speeds with shoes and barefoot are given in Fig.7 and Fig.8 respectively. The background color coding in the Figures 7, 8 coincide with the color coding for the gait phases described in Figure 6. The stance phase is detected when any of the sensory patches detect a force. The phase is considered finished when, typically, the anterior patch does not sense any force anymore and therefore the swing phase begins. In walking speeds (under 8 km/h), the stance phases of the feet overlap. However during running speeds (above 8 km/h), there are instances where neither socks sense any force, therefore correctly identifying float phases. Thus, the socks are capable of precisely identifying the different gait phases of walking as well as running.

In barefoot locomotion, smaller pressure amplitudes are observed throughout all gait cycles compared to locomotion with shoes. In most cases, all three sensors are activated during the stance phase. However, especially at higher locomotion speeds in the barefoot situation, the heel sensor often remains inactive. This again coincides with common gait characteristics where a person relies increasingly more on the forefoot when running barefoot, which could be beneficial for low extremity joints. If all three sensors are activated strongly (especially with shoes) we observe the classic order of activation from heel to mid-stance to toe-off. In terms of symmetry, the pressure profiles for gait with shoes are highly symmetric for the feet, while when barefoot, clear asymmetries are visible, especially in terms of which parts of the feet strike the ground. Here, both the strength of activation, timing, and duration of sensor activation alter significantly between the feet. Furthermore, higher stride frequencies are also observed when barefoot at the same locomotion speeds, yet again coinciding with common gait characteristics [17].

It is clear, that the sensory sock data are in line with prior reported biomechanical characteristics of walking and running. Not only does this demonstrate the ability of the soft sensor to capture important characteristics such as timing and amplitude of the standard gait cycle, but also nuanced details such as missing heel strike and gait asymmetry when running barefoot. The relationship between gait speed and the gait cycle duration, as depicted in Fig. 9, is a key research finding. As expected, the gait cycle duration decreases with increasing speed, a well-established relationship in biomechanics. On the other hand, the floating phase duration increases with increasing running speed, providing further insight into the dynamics of the gait cycle. This understanding is crucial for interpreting the mechanics of running. Kyle et al. presented the difference between running barefoot and running with

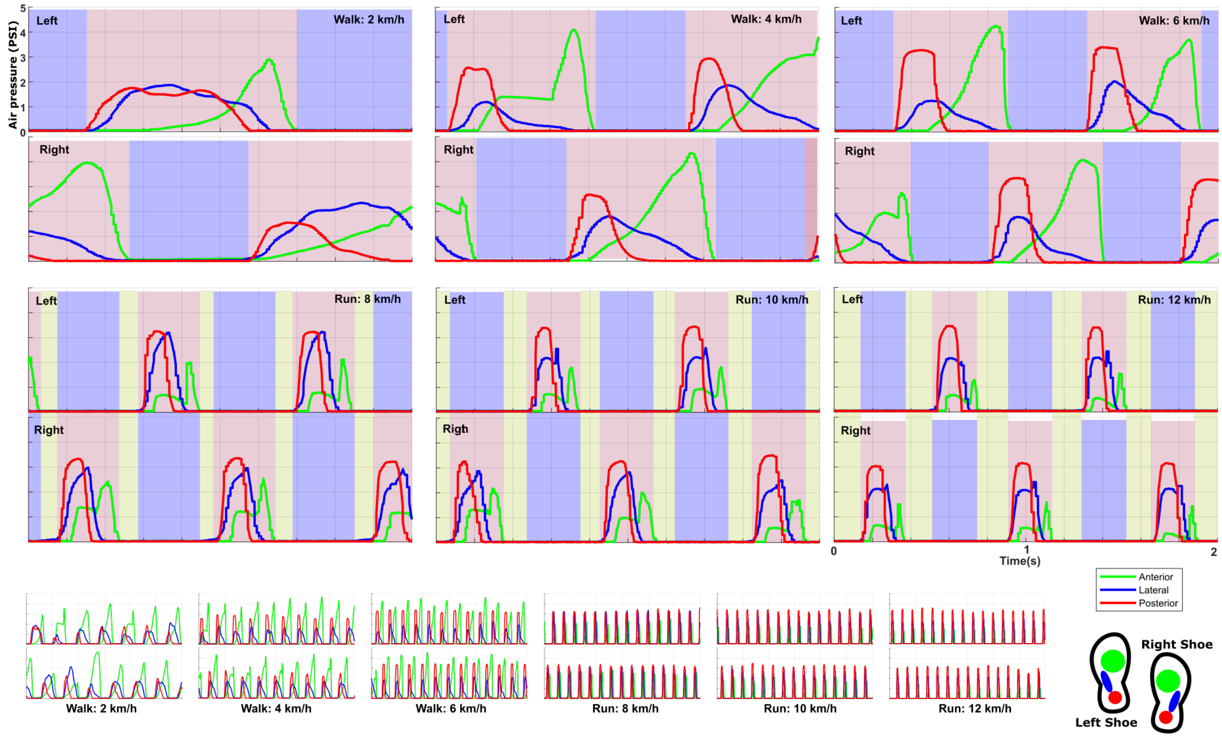

Fig. 7: Pressure sensors readings during walking and running at different speeds while wearing running shoes. The upper figures illustrate 2 seconds of the gait for each speed. The lower figures illustrate the readings during 12 seconds for each speed indicating the consistency of the sensory socks.

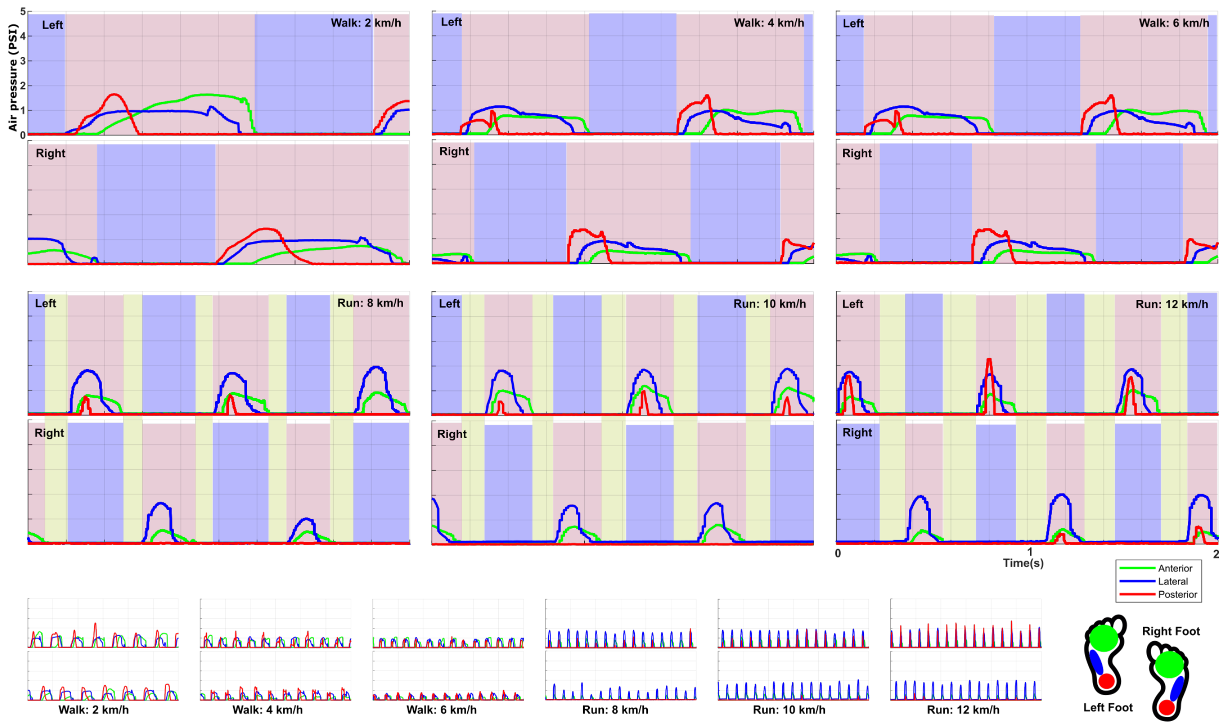

Fig. 8: Pressure sensors readings during barefoot walking and running at different speeds. The upper figures illustrate 2 seconds of the gait for each speed. The lower figures illustrate the readings during 12 seconds for each speed indicating the consistency of the sensory socks.

shoes in a review paper [14]. Interestingly, our proposed sensory system shows that in barefoot conditions, an overall less maximum vertical ground reaction force is observed; this result is consistent with the findings in [14].

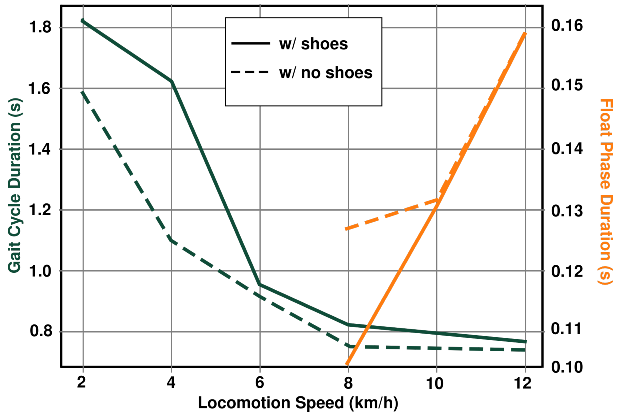

Fig. 9: Gait duration and floating phase duration at different walking and running speeds with and without shoes based on a single user.

V. Conclusion

The paper proposes a new design of soft sensory socks that measure ground reaction forces during gait while wearing shoes and also barefoot. Results on gait analysis performed on a single user with the sensory socks show the difference in gait while walking and running barefoot compared to with shoes. Furthermore, the proposed system is able to accurately detect the human gait cycle from very slow speeds (2 km/h) all the way up to fast speeds (12 km/h). These results indicate that the proposed soft sensory socks can potentially be used to enable the control of wearable lower body exoskeletons through precise foot sole pressure readings. Future work will address a larger number of study participants and a gait cycle phase predictor.

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