Ultra Thin Nanocomposite In-Sole Pressure Sensor Matrix for Gait Analysis

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Abstract Gait analysis plays an important role in various applications such as health care, clinical rehabilitation, sport training and pedestrian navigation. In order to monitor the human gait, an interesting approach is to analyze the foot plantar pressure distribution between the foot and the ground. In recent years, the emergence of flexible, soft and lightweight sensors facilitates the rapid technological advances in in-shoe foot pressure measurements, thereby especially carbon nanotubes-based sensors provide an outstanding solution for the implementation of flexible, soft pressure sensors in foot pressure distribution analysis. This chapter focuses on the design and implementation of multiwalled carbon nanotubes (CNT)/polydimethylsil-oxane (PDMS) based nanocomposite pressure sensors for the analysis of the foot pressure distribution. The sensor is durable, stable and shows sensitivity of 3.3 kΩ/kPa and hysteresis smaller than 3.64% with maximum detectable pressure up to 217 kPa, which is suitable for the measurement of human foot pressure. The proposed sensor has been implemented in a flexible in-sole, which is designed based on normal arch foot anatomy. A total of 12 sensors are distributed in the heel, lateral back foot, midfoot and front foot. The foot pressure distribution for different persons while walking and standing using nanocomposite sensor based in-sole were investigated by measuring the changing in resistance of the pressure sensors, when pressure applied on it. It shows that foot pressure distribution is higher in the fore foot and the heel while person standing in normal position. While walking, initially the foot pressure is in

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O. Kanoun and N. Derbel (eds.), *Advanced Sensors for Biomedical Applications*,
Smart Sensors, Measurement and Instrumentation 38,
https://doi.org/10.1007/978-3-030-71225-9_2

the heel and then transferred to the entire foot and finally it is concentrated on the fore foot.

Keywords Carbon nanotubes · Polydimethylsiloxane · Nanocomposites · Pressure sensors · Screen printing · Solution mixing · Gait analysis

1 Introduction

The foot is important for humans for the interaction with ground for standing and locomotion. According to a institute of preventive foot health, an average human walks about 160,000 km, which is equal to 4 times around the earth (Franklin et al. 2015). In last decades, humans are subjected to various foot problems such as bunions, corns and calluses, hammertoes, heel pain, arch problems, chronic foot pain and diabetes related wounds (Crawford et al. 2020; Jeffcoate and Harding 2003). According to the International Diabetic Federation, 327 million people in age groups of 20–64 around the world and 58 million people in Europe were diagnosed for diabetes and it will be increased to 438 million people in 2045 (Federation 2020). Indeed, the diabetes mellitus accounts for over \$1 billion per year in medical expenses (Federation 2020). Furthermore, gait instability in the elderly and other balance impaired individuals show the need of ways to analyze foot pressure distribution to obtain gait balance improvements which is considered important both in sports and biomedical applications such as forefoot loading during running, soccer balance training and foot balancing during weightlifting (Walther et al. 2020). It is estimated that around 13–59% people are subjected to foot injury during their daily activities (Walther et al. 2020).

In this chapter after a literature review on gait analysis, we introduce a low cost, highly durable, flexible binary nanocomposite consists of polydimethylsiloxane (PDMS) and multiwalled carbon nanotubes (MWCNT), which can reach high sensitivity and wide detectable pressure range. These sensors were implemented in a flexible in-sole where the design and sensor placement is based on the normal arch foot anatomy. This chapter concludes with evaluation of the in-sole by analysing the foot pressure distribution on a individual during walking and standing.

2 Gait Analysis

Gait analysis is important for providing the feet pressure distribution data to the physicians and therapists, to diagnose foot problems and walking disorders in order to find suitable treatments to patients and improving the gait stability during their sports and daily activities. To study the locomotion and kinematics of the foot, various models have been adopted to represent the gait pattern. In early stage, two dimensional models were considered to represent the foot. These models considered the foot as rigid body. Although the foot is a much more complicated structure with many individual muscle layers, bones, and joints (Lyons et al. 2006). During 1940s and 1950s, gait analysis was studied with the muscle activities during the different phase of the gait cycle by free-body diagrams and calculations on the effect of knee, hip and ankle joints (Lyons et al. 2006). In 1960s, many researchers employed mathematical modeling to demonstrate the motion of the body segments and actions of different muscles (Abdul Razak et al. 2012). Later, kinetograph, which is an apparatus for taking a series of photographs of moving objects for examination with the kinetoscope was introduced to measure the foot pressure distribution during walking motion in low cost measurement systems. The obtained kinetograph was analyzed by x-ray images and can be improved by placing a black rubber mat with reflecting pyramidal projecting fluid on a glass plate (Lyons et al. 2006). Further advances in gait analysis came with introduction of more accurate kinematic instruments in 1980s, which results in improved kinematic studies using electronics rather than images or visual observations that took a long time to gather information and also force platforms and EMED systems were made available, which produced reliable results in minutes (Zulkifli and Loh 2020). However, these advances in gait analysis are still expensive, as it requires dedicated equipment for the motion capture. Late 1990s, foot pressure analysis was enhanced using EMED sys-tem, F-scan systems, PEL-38 electronic podometer and piezoelectric sensors, which was placed under the foot and motion analysis camera has been used to investigate the gait pattern (Zulkifli and Loh 2020). However these systems were limited to the static foot pressure measurement.

In 21st century, various low-cost conventional techniques such as silicone layer, air bladder and optical fibers were used to enhance the foot pressure measurement in both static and dynamic mode (Kong and Tomizuka 2008; Soetanto et al. 2011). However, these techniques are not promising for the large deformation and repeating pressure cycle because of its limited mechanical strength. Later newer technology enabled electrical sensors to be implemented in the foot pressure measurement in two methods: platform and in-shoe. Platform systems are constructed from a flat, rigid array of pressure sensing elements arranged in a matrix configuration and embedded in the floor to allow and follow normal gait. Both static and dynamic measurement can be done in this system. But this system was restricted to the research laboratories and lack of on-field implementation (Abdul Razak et al. 2012). In-shoe sensors are flexible and embedded in the commercial shoes such that measurements reflect the interface between the foot and the shoe (Abdul Razak et al. 2012). Commercial sensors such as "Novel, Parotec, Tekscan, Vista Medical, Wahab" adopted both measurement systems to enhance the quality of the foot pressure measurement, where they can reach pressure range of 260 kPa–1.034 MPa and hysteresis of 0.05%–24% (Abdul Razak et al. 2012). The evolution of stretchable and flexible sensors paved the way for the many research works on the foot pressure analysis (Lou et al. 2017; Nobeshima et al. 2016; Pyo et al. 2017). Stretchable sensors based on conductive rubber, and conductive nanoparticles as nanofillers in soft polymers provides the flexible, durable and stable foot pressure measurement. These sensors have different layers, which act as artificial skin and it can reach sensitivity ranges from 0.1 W/N to 670 kW/N and detectable pressure from 3 to 600 kPa (Lou et al. 2017; Nobeshima et al. 2016; Pyo et al. 2017). To enhance the comfortability for the foot pressure measurement systems, these stretchable sensors are implemented in textiles "E-Textile" made of multiple layers such as carbon nano tubes coated polyester, graphene and organic polymers, which is suitable for the pressure range 0.01–1.2 MPa (Lin et al. 2016; Lou et al. 2017). Despite its comfort, it is not durable for long time loading application. In most of the works based on stretchable sensors employing MWCNT as nanofillers, one of the best choice for matrix are soft polymers Thermoplastic poly urethane, polyester, polyethylene terephthalate and PDMS (Polydimethylsiloxane). This is due to that the soft polymer has low Young's modulus can retain to its orginal form faster than other polymers and it can be utilized for dynamic and high frequency foot pressure applications like walking and running (Canavese et al. 2014; Cheng et al. 2011; da Costa and Choi 2017; Huang et al. 2017; Karimov et al. 2015; Lee and Choi 2008; Ramalingame et al. 2017a; Sepulveda et al. 2011; Shu et al. 2010; So et al. 2013).

3 CNT/Polymer Pressure Sensor

Recently, various research works on CNT-based sensors, have been highly active in interest for diverse applications. Such composites are obtained by introducing enough dispersed CNTs into a polymer matrix that enables sensing capabilities in the resulting nanocomposite. The conductivity and thus, the sensing properties of these composites depend on numerous factors, such as the quality and kind of the polymer matrix and the size, nature and concentration of the dispersed CNTs. When compressive forces are transferred to the surface of these nanocomposites, the distributed conductive particles are induced to contact each other, resulting in formation of more conducting paths than the already existing before applying the pressure and hence reducing the electrical resistivity of the nanocomposite (Fig. 1).

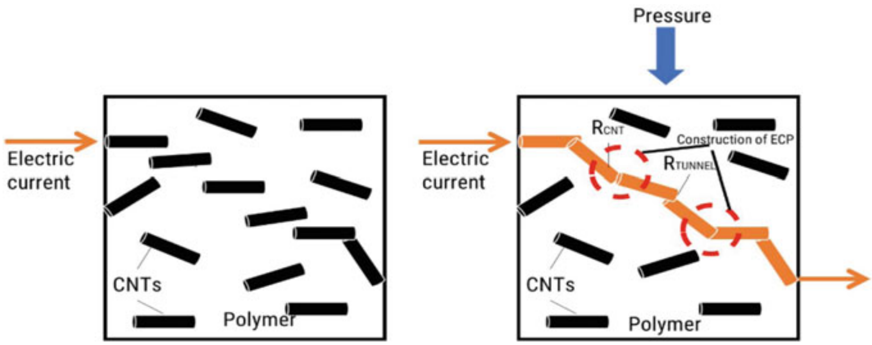

Fig. 1 Resistve principle of CNT/polymer nanocomposite pressure sensor

CNTs have been already incorporated in various polymer such as poly (methyl methacrylate) (PMMA), polycarbonate (PC), poly(L-lactide) (PLLA), pol vinyl alcohol (PVA) and others, in order to fabricate pressure sensors (Sousa et al. 2015). Among other nanocomposites, medical grade polydimethylsiloxane (PDMS) added with CNTs results in a nanocomposite material with high biocompatibility and flexibility, which suits pressure sensing application involving direct human contact (Maddipatla et al. 2017).

3.1 Fabrication Process of CNT/PDMS Pressure Sensors

Solution mixing approach is used to fabricate the CNT/PDMS nanocomposites, which involves sonication and mechanical stirring. These methods provide adequate stress to de-bundle the CNTs and disperse it in the PDMS effectively (Ramalingame et al. 2019). Tetrahydrofurane (THF) is used as common organic solvents, because it is one of the optimum solvent to disperse the CNTs efficiently (Ramalingame et al. 2017b). 0.3 wt. % MWCNT (Sigma-Aldrich, O. D×L—4.5 nm ×0.5 nm ±3–6 µm, 95%) with THF and sonicated using Bandelin sonoplus for 20% of the total power for 15 min and then stirred magnetically using CAT-M27 for 60 min, then mixed with PDMS (CNT/THF:PDMS—1:1).

The mixture is sonicated for 30 min at 50% amplitude and then stirred magnetically for 60 min at 70 °C followed by sonication of 50% amplitude for 30 min (Fig. 2).

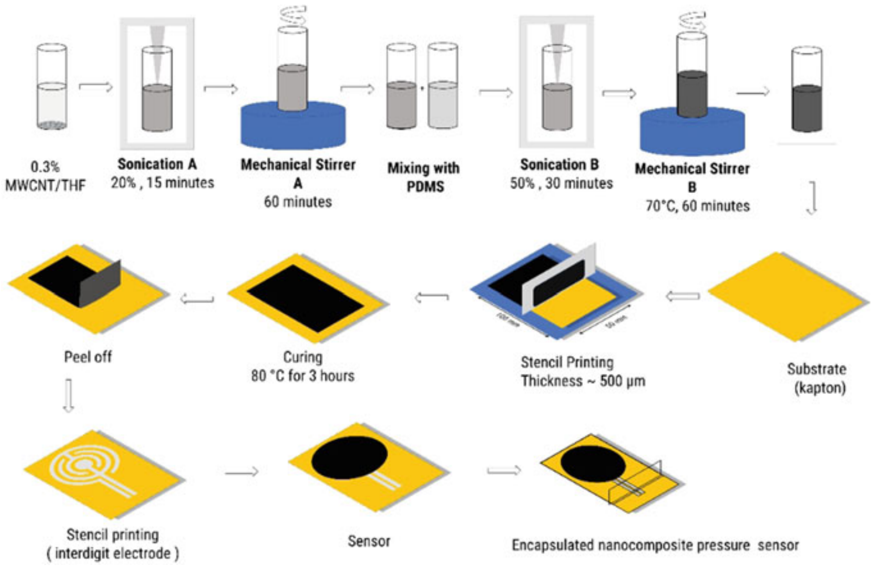

Fig. 2 Fabrication of CNT/PDMS nanocomposite pressure sensor

Curing agent is added at the ratio of 10:1 and mixed manually for 15 min to mix hardener uniformly in nanocomposite. Then, thin film is deposited over the Kapton substrate using stencils printing with thickness of 200 μm in the rate 1 mm/s. Then it is allowing for curing for 80 °C for 4 h. Once it is cured its peeled off from the Kapton and diced in 10 mm radius. In parallel, silver based interdigit electrode is printed over Kapton using stencil printing at the rate of 1 mm/s and cured at room temperature for 4 h. Then thin film is placed over the electrode and encapsulated using PET with the help of commercially available laminator.

The electrical conductivity of the MWCNT/PDMS nanocomposite pressure sensor is measured using a digital multimeter Agilent 24401A. The conduction mechanism is based on the percolation theory and tunnelling effect where CNTs act like metal-metal junctions form conduction paths between the electrode and polymer will act like tunnelling barrier between CNTs. This phenomenon can be visualized as three-dimensional resistors networks based on tunnelling effect (Kanoun et al. 2021). Theoretically, two types of resistance can be seen in MWCNTs-PDMS nanocomposite films such as R_{tube} is intrinsic resistance of CNT, which will be around 0.2 $k\Omega/\mu m$ to 0.4 $k\Omega/\mu m$ (Kanoun et al. 2021). The second type is Rjunction, which is further divided into contact resistance between CNTs (R_c) and tunnelling resistance (R_T) between CNTs, which can be seen below:

$$R = R_{tube} + R_{junction} \tag{1}$$

$$R_T = \frac{h^2 d}{Ae^2\sqrt{2m\lambda}} e^{\frac{4\pi d}{h}\sqrt{2m\lambda}} \tag{2}$$

where d is the distance between CNT, h is the Planck constant, e is the quantum of electricity, l is the barrier height of energy, m is the electron mass and A is the cross-sectional area of the tunnel (Sepulveda et al. 2011). During initial applied force of 1N, the nanocomposite sensor comes in contact with the underlaying electrode, which results in abrupt decrease in resistance of the sensor. So, the sensor needs an activation force of 1N to provide a stable measurement. Further increases in force results in further decreases in resistance.

This behaviour can be explained by Eqs. (1) and (2), where the change in orientation of conduction path results in change of R_c, change in tunnelling distance results in change of R_T, which in turn change the $R_{junction}$ and deformation of CNTs change in R_{tube}, which results in overall resistance decreases (Kanoun et al. 2021). From Fig. 3 it can be seen that the change in resistance gradually reduced at high force ($< \approx 100$ kPa), which is due to less formation of conductive paths and once it reaches a saturation point, there is no formation of new conductive paths (Kanoun et al. 2014). At this point, the change in resistance at high pressure region is due to the intrinsic resistance of CNTs, being slightly compressed.

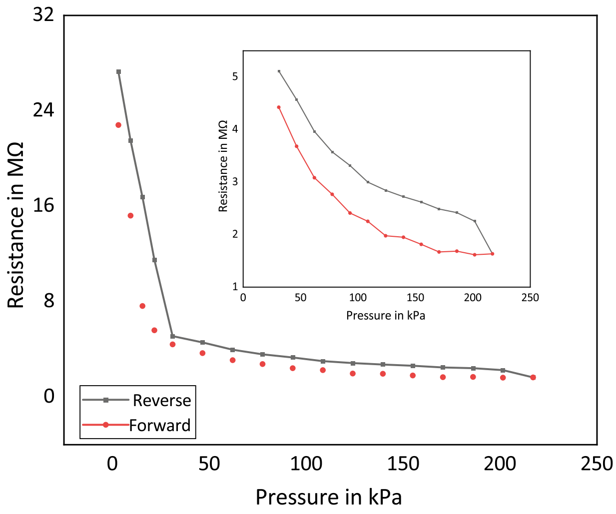

Fig. 3 Sensor behaviour for forward and reverse force cycle. Inside: Sensor behaviour for force of 3–217 kPa

4 Implementation and Evaluation

In-sole design and sensor placement are mainly dependent on foot anatomy like flat or high arched foot, clubfoot and an extra toe. Normally, a person is allowed to walk over the PressureStat film during a normal stride where these films will give the exact replica of the person footprint and sensors are placed in maximum pressure bearing points, which is reflected as darken area in Pressurestat film (Stalin 2012). In the work reported here, in-sole design is based on normal arch foot anatomy (UK size 10) and sensor is placed based on the pressure distribution for that foot taken from foot anatomy. A total of 12 sensors are distributed in the heel, lateral back foot, midfoot and foot. In heel, three sensors are placed. Pressure distribution is less in mid foot and so one sensor is placed. During the gait, the last phase of 1st stance will be on the forefoot and so seven sensors are placed in this region to get more resolution of pressure distribution in this region. The electrode layout is designed in adobe illustrator 2020. Then the electrodes are fabricated by silver inkjet printing using Diamtrix DMP-2850 obtaining 12 interdigitated electrodes plus a common ground. Next the nanocomposite films were prepared using fabrication technique explained above are placed over the electrodes and finally encapsulated with PET. Figure 4 shows the electrodes layout and sensors placement on the in-sole; the sensors were numbered as S_1 to S_{12} according to their location in the in-sole.

To test and validate our system, we first implement a measuring system with the help of Arduino microcontroller and a voltage divider circuit. A young adult subject participated in the controlled experiment to validate the system. Pressure distribution on the foot were analysed in stationary and dynamic phase. During stationary phase,

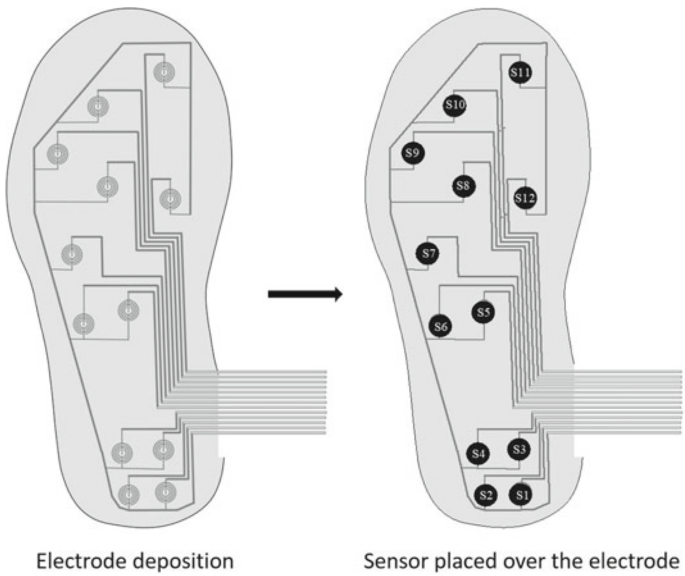

Fig. 4 Electrodes layout and sensors placement on the in-sole

the in-sole was placed on the ground and the subject could stand on it. During the dynamic phase, the insole was placed in each of the subject's shoes, and the subject was asked to perform different tests. Foot pressure data was recorded at time interval of 50 ms in the excel file and then extracted from the database after the experiment for further analysis.

4.1 During Stationary Phase

The aim of this experiment was to investigate the pressure distribution of the different sections of the foot such as forefoot, mid foot and heel at the stationary scenario. The subject was asked to do this test of each position for 7 s and 5 s gap in-between the different positions to ensure reliable and accurate results.

Figure 5 shows the maximum change in relative resistance of the pressure sensors placed on the in-sole when the person is asked to stand on it in different stationary position such as initial contact, loading response and heel off, all the positions are performed on the other leg support. Higher change in relative resistance means that more pressure is applied on the respective position. In initial contact position, the change in relative resistance of the sensors in the heel position are −79% to −86%. This is because in initial contact position, only the heel is in contact with the ground (in-sole) and most of the pressure will be applied on the heel position.

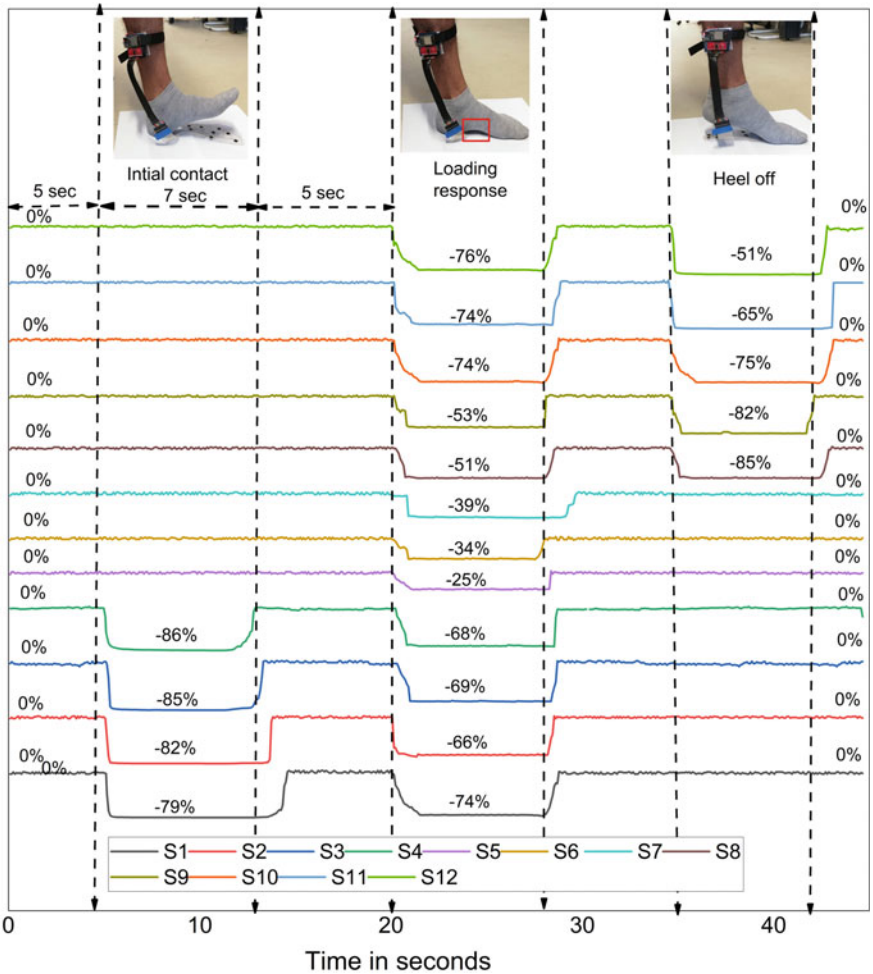

Fig. 5 Varying foot level like walking in stationary phase

4.2 During Dynamic Phase

The aim of this experiment was to investigate the dynamic gait pattern on a single leg during normal walking, which is considered as swinging leg and other one is the supporting leg. The gait pattern is comprised by two stances, each stance has five steps such as initial contact, load response, mid distance, terminal distance and pre-swing. In this work, the signal is recorded for the 1st stance of the walking (Fig. 6).

Figure 7 shows that maximum change in relative resistance of pressure sensors placed on the in-sole when the person subjected to normal walking. The first step of

Fig. 6 Evaluation of electronic in-sole in dynamic phase

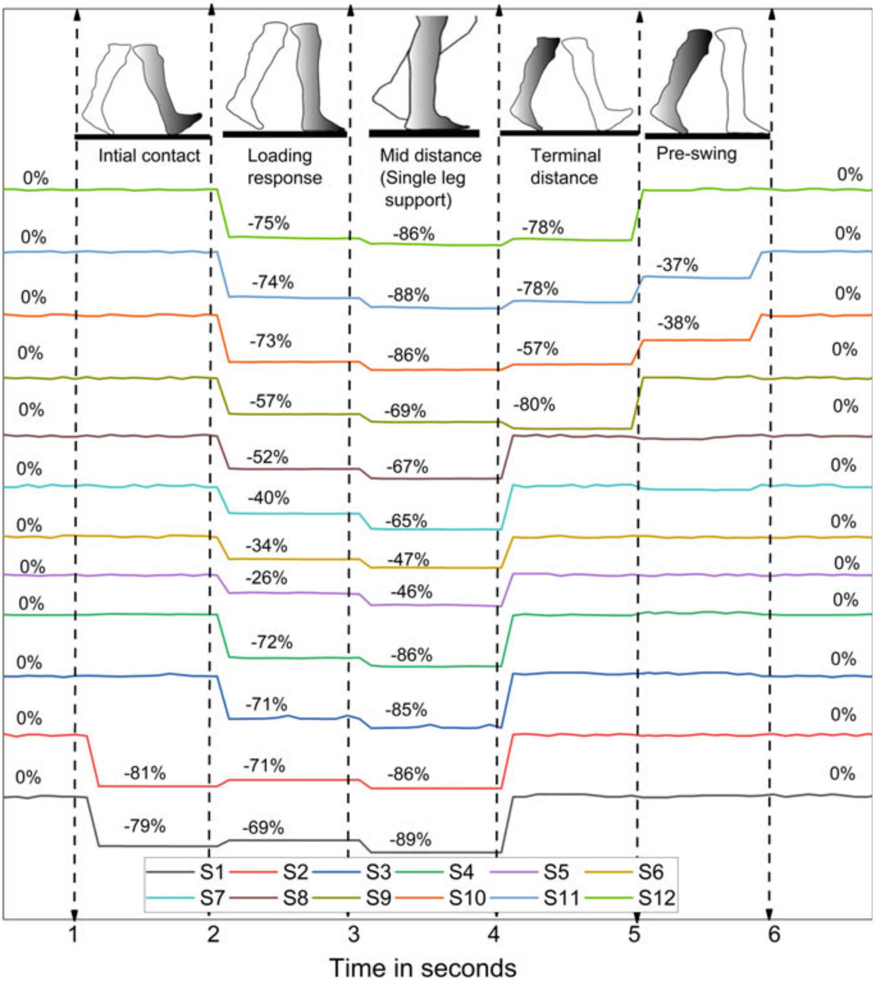

Fig. 7 Foot pressure distribution in different dynamic phase

1st instance of normal walking is initial contact where the body weight is completely focused on the heel of the swinging leg and so more pressure is applied on it. The next instance is when the body weight is transferred and distributed from the heel to the mid and fore foot of swinging leg. Despite of the pressure distribution, more pressure is applied on the fore foot and the heel. When the swinging leg reaches mid distance position, the respective foot is subjected to more pressure because the body weight is transferred from the supporting foot while it is not in contact with the ground. This can be seen from Fig. 7 that the change in relative resistance of the swinging foot is increased from in the range of 10–15% from the previous position. In terminal distance, the heel of the supporting foot and fore foot of the swinging foot will come to contact with the ground.

The body weight is transferred on both feet and so the pressure applied on the fore foot of the swinging leg is reduced compared to the mid distance position. This can be seen as, the change in relative resistance of the swinging leg is re-duced further in the range of 35%–45%.

5 Conclusion

In this chapter, after a brief analysis on various methods for monitoring the gait pattern and foot pressure distribution, we introduced pressure sensor based on Multiwalled carbon nanotubes (MWCNT) and polydimethylsiloxane (PDMS) with good sensitivity and wide detectable pressure range for gait analysis. We reported the design and fabrication technique of nanocomposites and pressure sensor based on CNT polymer nanocomposites in order to achieve a low cost, highly durable and highly sensitive sensing in-sole. The proposed sensors show sensitivity of $3.3\,\mathrm{k\Omega/kPa}$ in the range of 110 kPa and hysteresis of $<3.64\%$ with maximum detectable pressure up to 217 kPa. This chapter concluded with analysis of the foot pressure distribution while the person standing in stationary phase and gait pattern of the person during walking.

Acknowledgements The research work was carried out under the project “SenseCare—high-tech sensor technology for the challenge of demographic change in Saxony (100270070)” funded by the European Social Fund and the Free State of Saxony, “Landesinnovationsstipendium (100284169)”, funded by the Sächsische Aufbaubank (SAB) and the European Social Fund (ESF) and by the Deutsche Forschungsgemeinschaft (DFG, German Research Foundation)—Project-ID 416228727—SFB 1410, applicant Prof. Dr.-Ing. Olfa Kanoun.

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