In-Shoe Plantar Pressure Measurement and Analysis System Based on Fabric Pressure Sensing Array

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***Abstract*—Spatial and temporal plantar pressure distributions are important and useful measures in footwear evaluation, athletic training, clinical gait analysis, and pathology foot diagnosis. However, present plantar pressure measurement and analysis systems are more or less uncomfortable to wear and expensive. This paper presents an in-shoe plantar pressure measurement and analysis system based on a textile fabric sensor array, which is soft, light, and has a high-pressure sensitivity and a long service life. The sensors are connected with a soft polymeric board through conductive yarns and integrated into an insole. A stable data acquisition system interfaces with the insole, wirelessly transmits the acquired data to remote receiver through Bluetooth path. Three configuration modes are incorporated to gain connection with desktop, laptop, or smart phone, which can be configured to comfortably work in research laboratories, clinics, sport ground, and other outdoor environments. A real-time display and analysis software is presented to calculate parameters such as mean pressure, peak pressure, center of pressure (COP), and shift speed of COP. Experimental results show that this system has stable performance in both static and dynamic measurements.**

***Index Terms*—Fabric pressure sensor array, plantar pressure, wireless communication.**

I. Introduction

PLANTAR pressure is the pressure on the foot skin that the human foot experiences during daily activities. The first and direct use is the evaluation of footwear. It was used [1] to determine the effectiveness of therapeutic and athletic shoes with and without viscoelastic insoles, using the mean peak plantar pressure as the parameter. Guidelines were given by Mueller for the application of plantar pressure assessment in the evaluation and design of footwear for people without impairments [2]. The second application is related to athletic training. Frederick and Hartner endeavored to realize their dreams of optimized sports performance with thin-film pressure sensors and relatively inexpensive data acquisition hardware [3]. After 2000, more researches have been reported on athletic plantar pressure analysis in order to improve sports achievements, such as soccer balance training [4] and forefoot loading during running [5]. Clinical gait analysis, the investigation of the pattern of walking, is the third application. Morris and Paradiso developed a shoe-integrated sensor system for wireless gait analysis and real-time feedback, in which spatial pressure distribution of the foot was used in pattern recognition and numerical analysis [6]. Plantar pressure distribution was employed to detect gait patterns: normal gait, toe in, toe out, oversupination, and heel walking gait abnormalities [7]. Plantar pressure was applied in pathology foot evaluation, e.g., flat foot diagnose [8], assessment of the diabetic foot [9], strephenopodia, and strephexopodia [10].

A variety of plantar pressure measurement systems are available in the market or in the research laboratories. With respect to their technical specifications and intended application, in general, there are two main types of devices: platform systems and in-shoe systems. Platform systems are usually embedded in a walkway. The typical commercial product is EMED®-SF floor-mounted capacitance transducer matrix platform (Novel USA, Inc., Minneapolis, MN), which is frequently used in clinical researches for diabetes mellitus [11], [12], gait analysis [13], hallux valgus [14], etc. However, this kind of system is restricted to use in a laboratory or hospital, and used for barefoot measurements. In-shoe systems can be used to record the plantar pressure distributions within a shoe. Commercial products include F-scan measurement system (Tekscan, Inc., South Boston, USA) and Novel pedar system (Novel USA, Inc.,) that capture dynamic in-shoe temporal and spatial pressure distributions, which were utilized for dynamic gait stability analysis [15], gait detection [16], and altered gait characteristics during running [17]. However, both systems use electrical wires to connect in-shoe sensors and data acquisition system around the waist, which cause inconvenience and discomfort during strenuous exercises. They are not suitable for long-term outdoor measurements. A wireless structure shoe-integrated sensor system was developed for gait analysis and real-time feedback [6]. In the system, hard devices have been used as the sensing units, which are not comfortable and cannot last for a long time because of fatigue of the sensing units. The data acquisition systems are often large and cannot be configured to connect with different remote receivers such as smart phone.

With the purpose of collecting valid long-period data in all kinds of activities, the plantar pressure measurement system should have the following features. First, it should be wearable

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to realize outdoor measurements. In this case, in-shoe systems are preferred than platform ones. Second, the pressure-sensing units should have sufficient accuracy and reliability under a large number of repeated loading cycles. Third, the equipment should not bring inconvenience and discomfort to the wearer in order to make sure the wearer is in a natural status as usual. It is the most important feature for getting real plantar pressure data without interferences. However, seldom plantar pressure measure systems accord with this feature. Fourth, wireless technology should be used to release the activity limit brought by wire connection. Based on the aforementioned four features, this paper presents an in-shoe plantar pressure measurement and analysis system based on a fabric pressure sensing array. A textile pressure sensing array developed by the authors' group is embedded in an insole. A data acquisition system is designed with small size and stable to power supply interference. Bluetooth virtual serial port technology is utilized based on which three flexible wireless configuration modes are developed to make this system suitable for research laboratories, clinics, and daily outdoor/indoor activities. User interfaces are presented to give a real-time display and analysis on spatial and temporal plantar pressure distributions.

The remainder of the paper is organized as follows. Section II describes the insole fabrication, including textile pressure sensing array, soft circuit, and insole package. Section III discusses the data acquisition and processing. A prototype of plantar pressure measurement and analysis system has been presented. Experimental results and conclusions are provided in Sections IV and V, respectively.

II. Fabrication of Insole

A. Textile Pressure Sensing Array

A family of soft pressure sensors has been recently developed by the authors' group using conductive textile fabric sensing elements. The pressure sensor [see Fig. 1(a)] was fabricated by adhering a conductive sensing fabric with conductive yarns and a top-and-bottom conversion layer, which were prepared by molding of silicon rubbers of varied Young's modulus beforehand. A knitted fabric coated with carbon-black-filled silicon was used as the strain sensing element. It has a strain gauge factor of approximately 10 or above and excellent fatigue resistance (>100 000 cycles) for strain up to 40%. The sensor measurement ranges are from 10 Pa to 800 kPa, suitable for a wide variety of human-apparel interfaces, such as loosely fit garment, pants for sitting, gloves, sleeping garments, and walking/running shoes. The accuracy is 5% and zero drift less than 5%. The sensor is packaged by silicon rubber so that moisture and dust will not affect its performance, as shown in Fig. 1(b). It is soft, which is essential for comfort consideration in daily activities. The second obvious advantage is its high sensitivity. Fig. 1(c) is a resistance–pressure curve of these textile pressure sensors, which is typically nonlinear, but with a reasonable repeatability. A higher sensitivity is helpful for the interface circuits to get a higher accuracy and resolution. The sensing fabric has been tested in 100 000-cycle loading test, and also has long service time, which is very essential for shoes, which

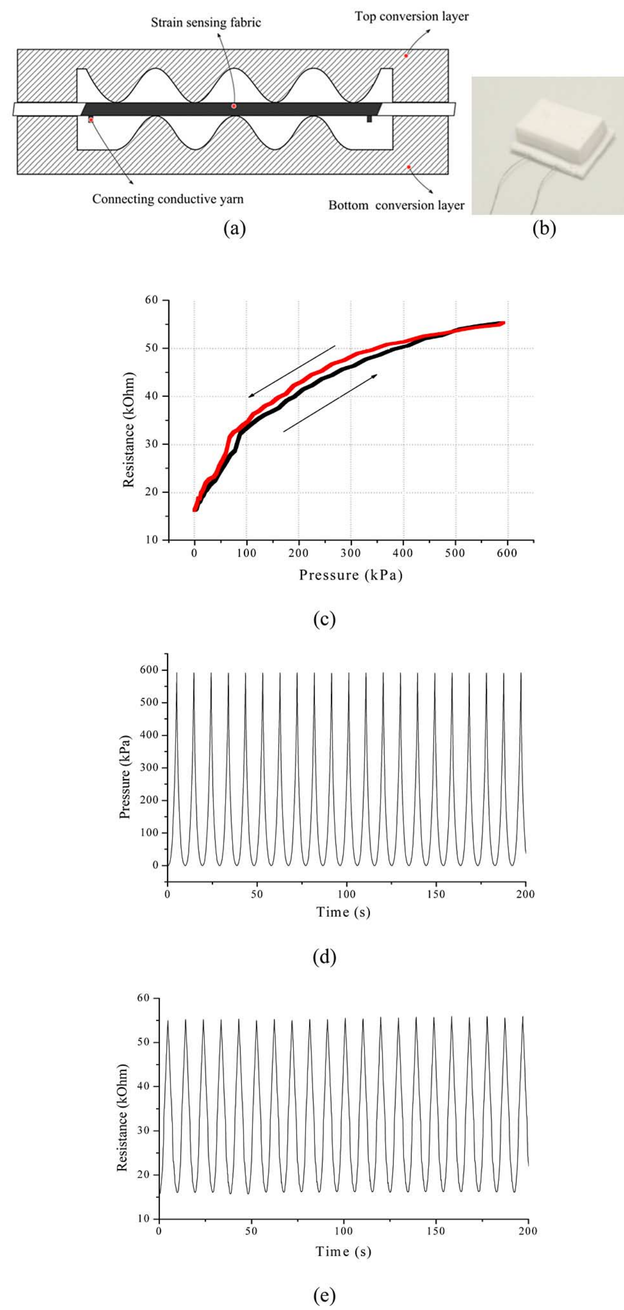


Fig. 1. Textile pressure sensor. (a) Schematic diagram of the structure of the pressure sensor (side view). (b) Package outlook. (c) Typical pressure resistance curve of fabric pressure sensor. (d) Pressure against time in a cyclic loading test. (e) Resistance against time in a cyclic loading test.

should be worn for a long period. Fig. 1(d) and (e) illustrates the responses (pressure and resistance) of a fabric pressure sensor plotted against time in a cyclic loading test. These textile pressure sensors can be also used in other wearable applications.

Several textile pressure sensors are connected in an "$n + 1$" line structure to construct a sensor array. One line is connected to each sensor as the "1" ground line of the sensor array. Another line of each sensor is the output as the "nth" signal line for the sensor array that contains n sensors. By combination of several textile pressure sensors, the sensor array is able to measure the

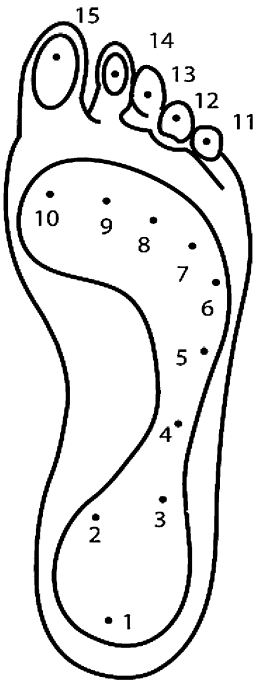
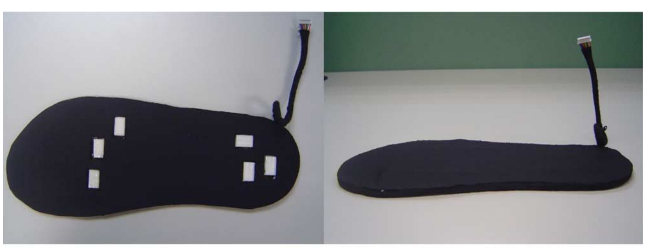


Fig. 2. Foot anatomical areas.



(a) (b)

Fig. 3. Insole and its package. (a) Textile sensors with bottom layer. (b) Insole.

pressure on a single-point region, as well as the distributions. In this prototype, a six-sensor array was used to construct a sensor array in "6+1" line structure.

B. Sensor Position Selection

The sole of foot can be divided into 15 areas, as heel (area 1–3), midfoot (area 4–5), metatarsal (area 6–10), and toe (area 11–15), as shown in Fig. 2. These areas support most of the body weight and adjust the body balance. The measured force at these positions can be used to derive physiological, structure, and function information of the lower limbs and whole body [18]. In order to reduce the system complexity, six positions (see Fig. 3) were selected at heel and metatarsal areas in the first prototype shoe, because these areas have higher pressure during normal activities of children, young, and old adults [19], [20]. Exact locations of six sensors were determined by depth shape of the subject's foot in soft model. Bamberg *et al.* [21] utilized five sensors at the positions of heel, metatarsal, and hallux to achieve a clinical gait analysis. Pappas *et al.* [22] presented a gait-phase detection system using three sensors at underneath the heel and metatarsal areas. Six sensors at the heel and metatarsal positions are adequate for such clinical investigation as gait analysis. For sports and fitness, the Nike+iPod Sport Kit used just one sensor in the midfoot to measure the wearer's pace, distance, and energy consumption during running [23]. The Adidas-1 running shoe uses one sensor at the heel position to provide compression measurement for adjusting to running situation [24]. Hence, six sensors are sufficient for sport or fitness assessments. Ideally, adding additional sensors at hallux and midfoot positions will enlarge the application scopes. As the textile sensor is relatively cheap and has a good fatigue resistance, it is easy to add more sensors in other applications.

C. Soft Circuit and Insole Package

A polyimide film circuit board was used to fabricate the sensing array in a foot shape. Six block positions, signal paths, ground paths, and connectors are lined out in the soft circuit. Six textile sensors are adhered at the corresponding positions of the soft circuit. Three are put in the forefoot and the other three are placed in the heel. Conductive yarns from each sensor are mechanically connected and adhered with the soft circuit, which is sandwiched by soft foam layers at the top and bottom. The half and whole integrated insole can be seen in Fig. 3(a) and (b), respectively. A data bus with a proper connector is adhered with the soft circuit and packaged together with the insole.

D. Temperature and Humidity Effect

The effects of temperature and humidity were measured on the sensing element, i.e., the fabric sensor. The resistance change caused by humidity change from 40% to 90% is about 2%, which is very small. Temperature compensation is necessary for an environment with significant temperature variation. However, as the insole temperature was almost constant in our investigation, the compensation was not incorporated.

III. Data Acquisition and Processing

A. Data Acquisition System

A wearable, wireless data acquisition system was developed as an electronic interface for the resistive textile sensors. Fig. 4 shows the data acquisition system diagram. Through the connector, voltage signals on the sensors are extracted from voltage dividers ($R_{\rm ref} = 30$ kΩ) and then sent to the embedded analog-to-digital (A/D) converter channels in the microcontroller PIC18F452. After quantization in the 10-bit A/D converter, the digital voltage values are transformed into resistance values. Through the serial port connection, resistance values arrive at Bluetooth module. Finally, these values are wirelessly transmitted to a remote receiver by the Bluetooth antenna. A voltage regulator is used to convert 3.7 V from a Li-ion battery to 3.3 V power supply as the system $V_{\rm cc}$. The data acquisition system has the following advantages.

1) Simple structure that leads to a small size and light weight.
2) Large working range and acceptable accuracy for wearable resistive sensor.
3) Stable performance to power supply interference.
4) Removable and rechargeable battery configuration.

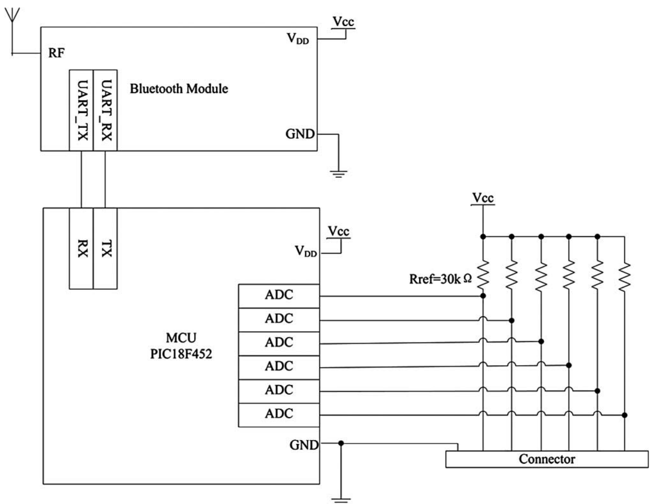


Fig. 4. Data acquisition system diagram.

The six textile pressure sensors in the insole array are connected to the $R_{\rm ref}$ and ground (GND) through the connector. Using the voltage division equation, we can get the A/D converter input analog voltage ($A_{\rm input}$) as

$$A_{\rm input} = \frac{R_{\rm ref}}{R_{\rm ref} + R_{\rm sensor}} V_{\rm cc}. \tag{1}$$

Output digital value ($D_{\rm output}$) can be obtained as [25]

$$D_{\rm output} = \left[2^m \frac{A_{\rm input}}{V_{\rm ref}} \right] \tag{2}$$

where $[\cdot]$ denotes the integer part of the argument, $V_{\rm ref}$ is the input full-scale voltage, and m is the bit length of $D_{\rm output}$.

The $V_{\rm ref}$ of embedded A/D converter channels is a software selectable to the microcontroller's positive supply voltage $V_{\rm cc}$. This configuration makes $V_{\rm ref} = V_{\rm cc}$. Then, (2) can be derived as

$$\begin{aligned} D_{\rm output} &= \left[2^m \frac{A_{\rm input}}{V_{\rm ref}} \right] = \left[2^m \frac{\left(R_{\rm ref} / \left(R_{\rm ref} + R_{\rm sensor}\right)\right) V_{\rm cc}}{V_{\rm cc}} \right] \\ &= \left[2^m \frac{R_{\rm ref}}{R_{\rm ref} + R_{\rm sensor}} \right]. \end{aligned} \tag{3}$$

Equation (3) shows the relationship between digital resistance values from embedded A/D converters and sensor resistances. $V_{\rm cc}$ does not affect the digital outputs. Hence, this data acquisition system can achieve a stable performance even if the power supply is unstable. This feature is especially helpful in wearable applications, because wearable situation in daily activities is likely to attract more noise. Connecting the data acquisition system and the insole, we have a prototype of the textile-sensor-based in-shoe plantar pressure measurement system, as shown in Fig. 5.

B. System Configuration Modes

Bluetooth is utilized as the wireless communication technology in the present system. The wearer is required, in the overlapped radiation range of both Bluetooth modules from the remote receiver and the data acquisition system, to ensure a stable and continuous wireless communication path between these two objects. It is a limitation that the wearer is restricted to a short distance range (<10 m) from the remote receiver. But the reason to choose Bluetooth technology lies in its extensive use and easy accessibility in consumer electronics, such as computer, mobile phone, etc.

Fig. 5. Prototype of the in-shoe plantar pressure measurement system.

Fig. 6. Configuration mode with smart phone (on the playground).

We have developed three system configuration modes from the in-shoe plantar pressure measurement system to the remote receiver, distinguished by three types of remote receivers: desktop, laptop, and smart phone. In the first configuration mode, the remote receiver is a desktop computer with a Bluetooth adapter. The wearer is restricted in the range of radiation from Bluetooth adapter. It is suitable for clinical or research laboratories, where the subject is measured within a small activity room. In the second configuration mode, the remote receiver is a laptop with an integrated Bluetooth module. The wearer is still restricted in the radiation range of laptop Bluetooth module. While the laptop can be comfortably taken in outdoor environment, this mode can be used for measurement in small-area outdoor activities. The third configuration mode is the most flexible one. The remote receiver is a smart phone with embedded Bluetooth module. We have designed a wristband to fix the smart phone; therefore, it can record the foot pressure of the wearer during the activity without any location restriction. It is suitable for long-time outdoor measurement. Fig. 6 demonstrates this configuration mode.

Mode 3 is the most flexible one although it may be defeated by mode 1 and 2 as far as the performance of plantar

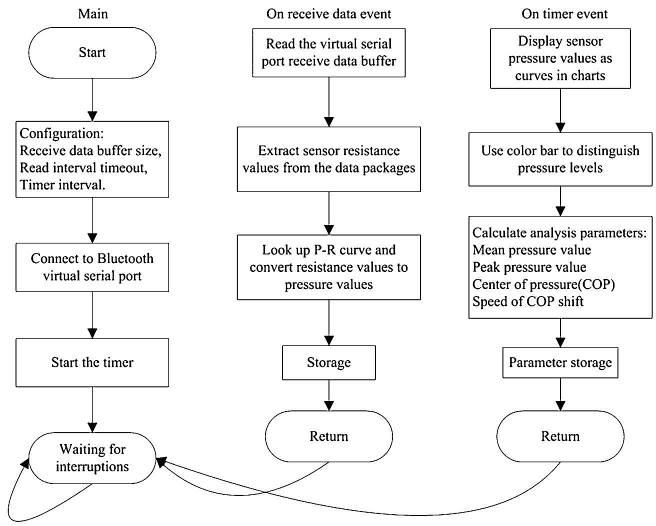


Fig. 7. Flowchart of data processing in the remote receiver.

pressure data processing and analysis is concerned. There are obvious hardware performance differences between the three types of remote receivers used in the prototypes. For example, we have used desktop Dell OptiPlex 755(CPU 2.66 GHz, RAM 2 GB, hard disk 160 GB), Laptop Lenovo ThinkPad X61(CPU 2.1 GHz, RAM 2 GB, hard disk 100 GB, Battery time about 5 h), and Smart phone HTC Touch Cruise (CPU 528 MHz, ROM 512 MB, RAM 256 MB, 8 G SD card, battery time about 6–8 h). Because of the hardware technology progress in smart phone, the hardware configuration of smart phone cannot be on par with desktop and laptop. It is mainly to record the plantar pressure data (its battery life time is sufficient for a normal outdoor activity record) and simple data processing. The desktop and laptop are able to run complex data processing and analysis.

C. Data Processing

In the remote receiver, data processing is done as shown in the flowchart of Fig. 7. The data acquisition system is manually connected to the remote receiver through wireless Bluetooth path, and is viewed as a Bluetooth virtual serial port. At first, a total configuration, including receiver data buffer size, read interval timeout, and timer interval, is set carefully and concertedly to ensure a following smooth data reception, display, and processing. Then, after connecting to the Bluetooth virtual serial port, the timer is forced to start with a following waiting status for interruptions. There are two event-driven responses, on receiving data event and timer event, corresponding to receiving data and processing data, respectively. On receiving data event, sensor resistance values are extracted from the data packages after reading out the virtual serial port data buffer, and transformed into pressure values according to pressure–resistance curve of the textile sensor [see Fig. 1(c)]. After this pressure value data storage, the program will return to the main body. On the timer event, these sensor pressure values are real-time displayed as moving curves in charts. Various colors are assigned to each sensor positions basing on a pressure level color bar.

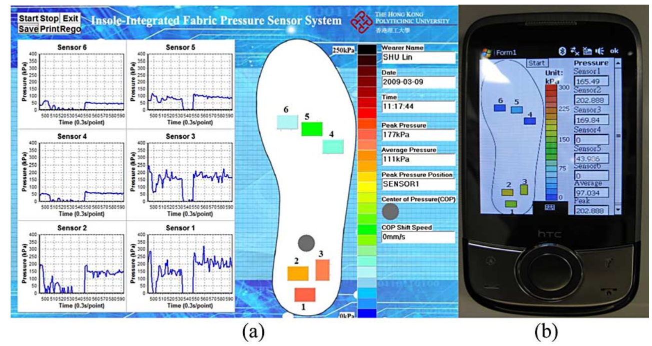


Fig. 8. Plantar pressure data interface for (a) desktop and laptop and (b) smart phone.

Some early analysis parameters, mean pressure, peak pressure, center of pressure (COP), and speed of COP, are calculated and stored sequentially. The program will also return to its main body. Parameters are calculated by the following equations:

$$\text{Mean} = \frac{1}{n}\sum_{i=1}^{n} P_i \tag{4}$$

$$\text{Peak} = \text{Max}(P_1, \ldots P_i, \ldots P_n) \tag{5}$$

$$X_{\text{COP}} = \frac{\sum_{i=1}^{n} X_i P_i}{\sum_{i=1}^{n} P_i} \qquad Y_{\text{COP}} = \frac{\sum_{i=1}^{n} Y_i P_i}{\sum_{i=1}^{n} P_i} \tag{6}$$

$$\begin{aligned}\text{Speed}_{\text{COP}} = \frac{u}{\Delta t}(&|X_{\text{COP}}(t+\Delta t) - X_{\text{COP}}(t)|^2 \\ &+ |Y_{\text{COP}}(t+\Delta t) - Y_{\text{COP}}(t)|^2)^{1/2}\end{aligned} \tag{7}$$

where n denotes the total number of sensors, i denotes a certain sensor, X, Y is the coordinate of the whole foot shape area, u is the unit distance between two neighbor coordinate points, and Δt denotes the time interval.

Two kinds of user interfaces are developed: one is for desktop and laptop users, as shown in Fig. 8(a), and the other is for smart phone users, as shown in Fig. 8(b). The programming tools are Delphi and Lazarus, respectively.

D. System Features

The technical features of the system are reported as: 100 Hz maximum sampling rate, synchronous sampling mode, 10–1000 kPa measurement range, 1 kPa resolution, and ±5% accuracy. As for the data compression, during data processing, display, and export for further use, the six-sensor pressure values are represented in a user-oriented format. Each sensor contains six bytes, including five bytes for integral and one byte blank for separation. One byte LG (line change) and one byte RE (return) are put at the end of a total round data. While in transmission, storage, and import, two bytes sampled value (SV) are used to represent each sensor in order to save transmission time and storage memory. User should import these data into the software to convert the sampled value to pressure display.

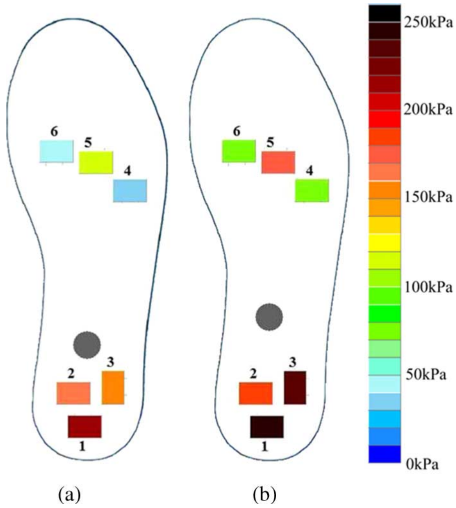


Fig. 9. Plantar pressure data displays: (a) two feet on the ground and (b) only the right foot on the ground.

IV. Experimental Results and Discussions

The experimental results presented here were obtained by using the aforementioned in-shoe plantar pressure measurement and analysis system based on textile sensor array. Fig. 9 illustrates a set of measurement results of a standing person (male and 72 kg weight). Fig. 9(a) is the right foot pressure when the wearer stands still with two feet supporting the whole body weight. The pressure readings of the right foot are lower, reflecting half of the body weight. Fig. 9(b) shows the pressure readings of the right foot when the wearer stands with his left foot in suspension. With just the right foot supporting the whole body weight, the pressure values of six sensors increase markedly. The black point, as COP moves a little to the middle of the foot, may reflect the wearer's balance control.

Dynamic tests, such as the walking test, were also conducted. The wearer was required to walk as usual, and the right foot pressure values are recorded in the laptop or the smart phone. Fig. 10 shows one set of the experimental records. Values of interest in this test sample are: the maximum mean pressure of the right foot is around 200 kPa; the maximum peak pressure of the right foot is about 300–350 kPa; the maximum mean pressure of the forefoot lies in the range of 100–120 kPa; the maximum mean pressure of heel is around 250 kPa; and the shift speed of COP is usually under 160 mm/s. These parameters are useful for clinical research, footwear design, and elderly wardship. Take elderly wardship, for example, the shift speed of COP exceeds a certain threshold, means that an elderly person is not in a stable, balanced walking status, which may be caused by musculoskeletal or neurological disorders. A foot pressure database can be set up to collect data from various people using the system over prolonged time period in normal activities.

A. COP Comparison

The COP results measured by our insole system were compared with those measured by the AMTI force platform (Advanced Mechanical Technology, Inc., MA). The prototype

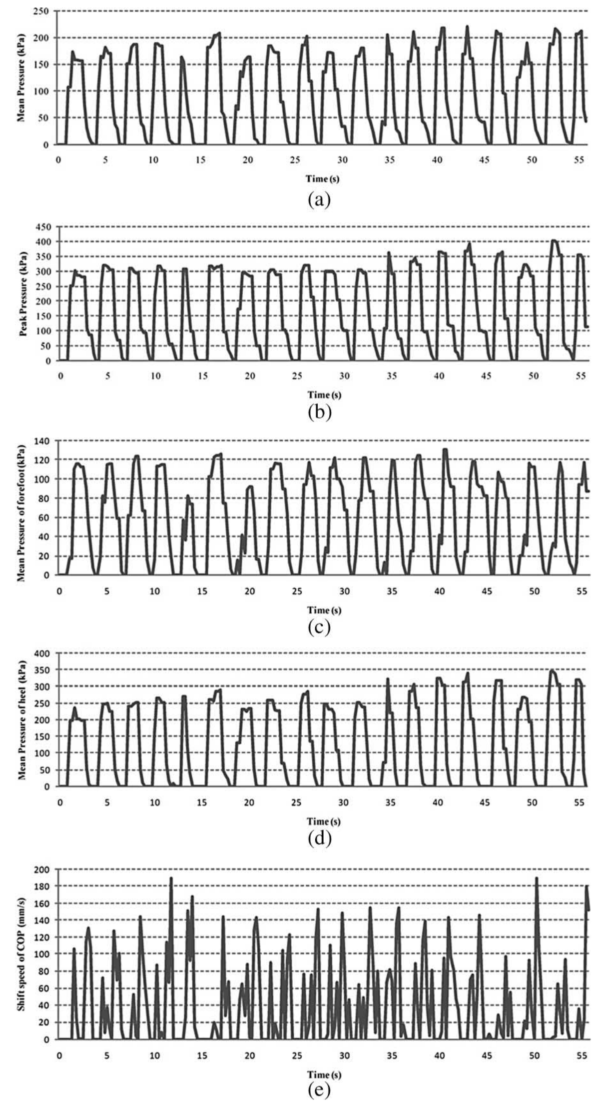


Fig. 10. Normal walking test (right foot). (a) Mean pressure. (b) Peak pressure. (c) Mean pressure of forefoot. (d) Mean pressure of heel. (e) Shift speed of COP.

TABLE I
Subject Information

Subject No.	Age(yrs)	Weight(kg)	Height(cm)	Foot size (European)
1	27	75	176	42
2	45	66	168	41
3	29	62	173	42
4	32	70	177	42
5	29	60	176	41
6	29	58	170	40
7	32	76	180	42
8	27	67	176	42
Mean(Std)	31.3(5.5)	66.8(6.2)	174.5(3.7)	41.5(0.7)

insoles were designed for male (foot size 42); hence, eight male subjects were chosen for the comparison experiments. Subjects' information can be seen in Table I. The insole coordinate was prematched with the force platform before it was fixed on the platform. Subjects were required to do four actions: normal standing, standing on one leg, heel strike, and push off, which are basic in normal activities. Fig. 11 shows the four actions

TABLE II
COP RELATIVE DIFFERENCE (IN PERCENT)

Subject No. (Times)	1(1)	1(2)	2(1)	2(2)	3(1)	3(2)	4(1)	4(2)	5(1)	5(2)	6(1)	6(2)	7(1)	7(2)	8(1)	8(2)	Mean (Std)	Total Mean(Std)
Normal standing	8.4	7.2	6.3	7.9	3.4	5.7	7.4	8.6	11.1	8.2	7.9	14.2	2.9	3.1	8.6	11.0	7.6 (2.9)	5.0 (4.6)
Standing on one leg	9.2	11.6	8.6	5.8	17.1	17.6	10.3	14.9	7.7	7.4	11.0	6.8	6.6	4.6	10.7	8.4	9.9 (3.7)	
Heel-strike	0.4	0.0	0.0	2.7	0.0	0.0	0.0	1.1	1.6	0.3	0.0	0.0	0.0	0.3	0.0	0.6	0.5 (0.7)	
Push-off	1.3	0.0	0.0	0.6	1.6	3.6	0.0	0.0	3.6	4.2	3.2	1.0	3.7	3.4	3.2	5.2	2.2 (1.7)	

Units: %.
COP reading was recorded as 1 × 1 cm square region; the difference will be marked as 0 if COP result from the force plate is inside the region.

TABLE III
WEIGHT DIFFERENCE IN VALIDATION TEST

Subject No.	1	2	3	4	5	6	7	8	9	10	Mean (Std)
Age(yrs)	32	32	29	63	27	27	29	27	29	29	32.4(10.3)
Height(cm)	177	180	173	170	176	168	170	176	175	176	174.1(3.6)
Foot size(European)	42	42	42	42	42	40	40	42	41	41	41.4(0.8)
Body weight(kg)	70	76	62	81	75	53	58	67	73	60	67.5(8.6)
Estimated weight(kg)	68	68	63	72	74	49	54	62	61	62	63.3(7.3)
Difference (%)	2.9	10.5	1.6	11.1	1.3	7.5	6.9	7.5	16.4	3.3	6.9(4.6)

Fig. 11. Test actions: normal standing, standing on one leg, heel strike, and push off.

recorded by a video recorder during test. Subjects should keep stable for 10 s at each action and tested for twice. The COP results were recorded by the force platform system and our system simultaneously. Relative differences between the two sets of recorded COP results are calculated and shown in Table II. Relative differences were calculated by dividing the diagonal length derived from landscape orientation with the longitudinal maximum lengths of the insole. Table II shows that the average difference is 5%, average differences during heel-strike and push-off tests are lower than 3%, while the differences in normal standing and one leg standing tests are higher. The results show that with these six pressure points, the measurement error in the location of the COP is low. Adding an additional sensor underneath hallux may further reduce the error. Our system was compared with Novel pedar-X system (Novel USA, Inc.) for their dynamic performance of COP measurement. The Novel pedar-X system has a sampling frequency around 78 Hz (total 256 sensors and 20000 sensors/second) [26], and our system has a maximum sampling frequency of 100 Hz, which is adequate to follow the variations of the COP during running or walking.

B. Weight Validation

A validation test was conducted to verify if the integration of the measured force on the feet surface gives a value that is close to the body weight of the subject. Ten male subjects were required to stand on one leg over the single insole and keep stable for about 10 s (the COP point should be kept within certain areas). Pressures of six sensors were recorded at the same time. An average pressure was calculated, and timed total area of six sensors, then divided by a weight factor (here, set 40%, which is an empirical value, meaning that the six sensors support 40% body weight). Finally, the body weight was estimated. The differences of estimated weight and real one are shown in Table III. While the values vary with individuals, the average difference is 6.9%.

V. CONCLUSION

In this paper, an innovative textile-sensor-based in-shoe plantar pressure measure and analysis system has been presented. Soft and sensitive textile pressure sensor, stable data acquisition system, flexible wireless configuration modes, and professional plantar pressure processing software have been developed, which makes this system suitable for research laboratory, clinic, outdoor test, and activity monitoring. For its low cost and easy to interface, ordinary people can afford to use in daily life. For example, one can use the system to test how much energy is consumed in an exercise, legs' coordination ability in running, balance control in activities, and comfort level with some certain plantar pressure. We have just test discrete pressures instead of the whole pressure distribution, in order to decrease system complexity, crosstalk error, and fabrication cost. The experimental results show that discrete pressures are able to reflect the wearer's status and plantar pressure change.

The results show that adding additional sensors at hallux and midfoot positions may improve the accuracy of COP measurement and weight estimation. In most situations, the temperature variation inside the shoe is small, so that there is no need for temperature compensation. However, temperature compensation will be required for applications where the temperature variation is greater than 10°C. The related investigations will be reported in a separate paper.

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REFERENCES

[1] L. A. Lavery, S. A. Vela, J. G. Fleishli, D. G. Armstrong, and D. C. Lavery, "Reducing plantar pressure in the neuropathic foot—A comparison of footwear," *Diabetes Care*, vol. 20, no. 11, pp. 1706–1710, 1997.

[2] M. J. Mueller, "Application of plantar pressure assessment in footwear and insert design," *J. Orthop. Sports Phys. Therapy*, vol. 29, no. 12, pp. 747–755, 1999.

[3] E. C. Frederick and K. P. Hartner, "The evolution of foot pressure measurements," *Sens. Mag.*, vol. 10, no. 6, pp. 30–35, 1993.

[4] A. Gioftsidou, P. Malliou, G. Pafis, A. Beneka, G. Godolias, and C. N. Maganaris, "The effects of soccer training and timing of balance training on balance ability," *Eur. J. Appl. Physiol.*, vol. 96, no. 6, pp. 659–664, 2006.

[5] R. M. Queen, B. B. Haynes, W. M. Hardaker, and W. E. Garrett, Jr., "Forefoot loading during 3 athletic tasks," *Amer. J. Sports Med.*, vol. 35, no. 4, pp. 630–636, 2007.

[6] S. J. Morris and J. A. Paradiso, "Shoe-integrated sensor system for wireless gait analysis and real-time feedback," in *Proc. 24th Annu. Conf. Annu. Fall Meet. Biomed. Eng. Soc. EMBS/BMES Conf.*, Houston, 2002, pp. 2468–2469.

[7] M. Chen, B. Huang, and Y. Xu, "Intelligent shoes for abnormal gait detection," in *Proc. IEEE Int. Conf. Robot. Autom.*, Pasadena, 2008, pp. 2019–2024.

[8] J. M. Tareco, N. H. Miller, B. A. Macwilliams, and J. D. Michelson, "Defining flatfoot," *Foot Ankle Int.*, vol. 20, no. 7, pp. 456–460, 1999.

[9] M. J. Mueller, M. Hastings, P. K. Commean, K. E. Smith, T. K. Pilgram, D. Robertson, and J. Johnson, "Forefoot structural predictors of plantar pressures during walking in people with diabetes and peripheral neuropathy," *J. Biomech.*, vol. 36, no. 7, pp. 1009–1017, 2003.

[10] B. Hodgson, L. Tis, S. Cobb, S. McCarthy, and E. Higbie, "The effect of 2 different custom-molded corrective orthotics on plantar pressure," *J. Sport Rehabil.*, vol. 15, no. 1, pp. 33–44, 2006.

[11] R. M. Stess, S. R. Jensen, and R. Mirmiran, "The role of dynamic plantar pressures in diabetic foot ulcers," *Diabetes Care*, vol. 20, no. 5, pp. 855–858, 1997.

[12] T. G. McPoil, W. Yamada, W. Smith, and M. Cornwall, "The distribution of plantar pressures in American Indians with diabetes mellitus," *J. Amer. Podiatric Med. Assoc.*, vol. 91, no. 6, pp. 280–287, 2001.

[13] U. Kanatli, H. Yetkin, and S. Bolukbasi, "Evaluation of the transverse metatarsal arch of the foot with gait analysis," *Arch. Orthop. Trauma Surg.*, vol. 123, no. 4, pp. 148–150, 2003.

[14] T. W. Kernozek and S. A. Sterriker, "Chevron (Austin) distal metatarsal osteotomy for hallux valgus: Comparison of pre- and post-surgical characteristics," *Foot Ankle Int.*, vol. 23, no. 6, pp. 503–508, 2002.

[15] E. D. Lemaire, A. Biswas, and J. Kofinan, "Plantar pressure parameters for dynamic gait stability analysis," in *Proc. 28th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. (EMBS)*, New York, 2006, pp. 4465–4468.

[16] P. Catalfamo, D. Moser, S. Ghoussayni, and D. Ewins, "Detection of gait events using an F-Scan in-shoe pressure measurement system," *Gait Posture*, vol. 28, no. 3, pp. 420–426, 2008.

[17] P. W. Kong and H. D. Heer, "Wearing the F-Scan mobile in-shoe pressure measurement system alters gait characteristics during running," *Gait Posture*, vol. 29, no. 1, pp. 143–145, 2009.

[18] Q. , H. Wei, W. , L. Lu, Z.Y. Fu, and S. B. Lu, "A new system for foot pressure measurement and gait analysis," *Chin. J. Biomed. Eng.*, vol. 19, no. 1, pp. 32–40, 2000.

[19] E. Kellis, "Plantar pressure distribution during barefoot standing, walking and landing in preschool boys," *Gait Posture*, vol. 14, no. 2, pp. 92–97, 2001.

[20] M. , J. Hessert, M. Vyas, J. Leach, K. Hu, L. , A. Lipsitz, and V. Novak, "Foot pressure distribution during walking in young and old adults," *BMC Geriatr.*, vol. 5, p. 8, 2005.

[21] S. Bamberg, A. Y. Benbasat, D. M. Scarborough, D. E. Krebs, and J. A. Paradiso, "Gait analysis using a shoe-integrated wireless sensor system," *IEEE Trans. Inf. Technol. Biomed.*, vol. 12, no. 4, pp. 413–423, Jul. 2008.

[22] I. P. I. Pappas, T. Keller, S. Mangold, M. R. Popovic, V. Dietz, and M. Morari, "A reliable gyroscope-based gait-phase detection sensor embedded in a shoe insole," *IEEE Sensors J.*, vol. 4, no. 2, pp. 268–274, Apr. 2004.

[23] T. Saponas, J. Lester, C. Hartung, and T. Kohno. (2006, Nov.). Devices that tell on you: The nike + ipod sport kit. Dept. Comput. Sci. Eng., Univ. Washington, Saint Louis, MO, Tech. Rep. [Online]. Available: http://www.cs.washington.edu/research/systems/privacy.html

[24] B. Eskofier, M. Oleson, C. DiBenedetto, and J. Hornegger, "Embedded surface classification in digital sports," *Pattern Recogn. Lett.*, vol. 30, pp. 1448–1456, 2009.

[25] B. Razavi, "Analog-to-digital converter architectures," in *Principles of Data Conversion System Design*. Piscataway, NJ: IEEE Press, 1995, pp. 96–152.

[26] (2009, Jun. 1). [Online]. Available: http://www.novel.de/productinfo/systems-pedar.htm

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