A Microprocessor-Based Data-Acquisition System for Measuring Plantar Pressures from Ambulatory Subjects

Hongsheng Zhu, Gerald F. Harris, Jacqueline J. Wertsch, Willis J. Tompkins, and John G. Webster

Abstract—**We have developed a portable microprocessor-based data-acquisition system to measure discrete plantar pressures within the shoe from ambulatory subjects. The system offers improved accuracy, repeatability, portability, and flexibility not available in current commercial systems. It consists of 14 conductive polymer pressure sensors, 14 analog amplifiers, an 8-bit analog-to-digital converter, a microprocessor, 120 kbytes of memory space, and a parallel I/O interface. Seven pressure sensors are embedded within each insole and located at the posterior heel, anterior heel, the four metatarsal heads, and hallux of each foot. The system is capable of continuously sampling 14 channels of pressure data for 7 min at a 20-Hz sample rate. The recorded data are downloaded into a microcomputer for further processing, analysis, and display. Foot pressures have been acquired from a sensate subject during multiple walking trials.**

Introduction

Diabetic patients often lose sensation in their feet. Lack of awareness of excessive plantar pressures may lead to neurotrophic ulceration [1]. As a result, various systems have been developed for measuring plantar pressures [2]–[15]. A classical method for measuring foot pressures is to employ a force plate or dynamometer system, typically embedded into a walking surface. Since the subject's foot must strike the plate surface, such systems may yield an unnatural walking pattern. Force plate systems generally measure composite pressure distributions of the plantar surface during foot contact. They are unable to measure discrete plantar pressures within the shoe and multiple plates are required to determine coupling effects between two feet during ambulation.

Placing sensors within an insole provides another method for measuring plantar pressures. This technique allows pressure quantification during the normal activities of a shoe-wearing subject. Discrete pressure sensors in the insole provide localized information, which can not be obtained from force plate systems. However, the available in-shoe pressure monitoring systems have thus far remained expensive with varying signal quality, lacking portability, and depending upon an umbilical connection to a remote computer for data-acquisition [2]–[14]. They typically have a limited data acquisition period (20 s) and allow recording of only a few consecutive steps. Such an early foot-to-ground force measuring device with an instrumented shoe was reported by Spolek and Lippert in 1976 [7]. The system was restricted to measuring heel and toe forces during several steps and was not portable. Miyazaki and Iwakura in 1978 presented a portable foot-force measuring system that used two pressure transducers under each foot [8]. In 1983 Franks *et al.* developed an umbilical microprocessor-based system which collected pressure data during a single step [9]. The Electrodynogram (EDG, Langer Biomechanics Group, New York, NY), a commercial system, employs a microprocessor for acquiring in-shoe foot pressures from seven plantar locations [10], [11]. The EDG can acquire data for 5 s from the 14 channels but has been reported to lack repeatability and accuracy [16]. The inherent variation was as great as 100–200% [16]. An umbilical system for measuring vertical reaction forces during gait was reported by Hermens *et al.* in 1986 in which eight capacitive sensors were attached to each sole [12]. The system could store 16 kbytes of data for several consecutive steps during a 20 s period. Capacitive sensors have also been used by Patel *et al.* (Hercules sensors, Allegheny Ballistic Lab, Rocker Center, WV) for measuring plantar pressures [13]. The Hercules sensors are relatively expensive ($150 each), thick (2.4 mm), and fragile [13]. The commercial EMED system (NOVEL Electronics, Minneapolis, MN) also uses capacitive sensors for pressure measurement and is relatively expensive ($6000 per insole). A portable strain-gage dosimeter to quantify foot-to-floor contact forces which stores 256 kbytes of data from six sensors mounted within a single shoe was described by Harris *et al.* in 1988 [14]. Although completely portable, accurate, and reliable, the system was expensive and limited to use by a restricted subject population. A recent commercial in-shoe pressure monitoring system, the F-Scan (Tekscan, Boston, MA) has been developed with 960 resistive sensors per insole. The insoles are usage-limited to a maximum of 50 cycles (steps) each and methods for calibrating the insoles which compensate for nonlinearity, bending, wear, and temperature drift are not described. The system utilizes an umbilical cable for data collection.

In this study, a portable microprocessor-based data-acquisition system was developed to monitor the pressure distribution under bony prominences of both feet during gait [15]. The fully portable system represents significant data acquisition improvements over previously described systems. The system continuously collects pressures between foot and shoe from 14 sensors for 7 min, which is much longer than the recording time of all previous systems. It provides discrete in-shoe plantar pressure information under seven locations of each foot. The data-acquisition unit is mounted within a 20 × 18 × 7 cm box and weighs about 0.8 kg. Subjects carry it in a backpack during ambulation. This system offers the advantages of portability and less interference to the subject's natural gait pattern. It enables the measurement of in-shoe foot pressures at discrete locations during various daily living activities.

Portable Pressure Data-Acquisition System

The portable microprocessor-based data-acquisition system consists of 14 conductive polymer pressure sensors, 14 amplifiers (LM 358), an 8-bit analog to digital converter with on-chip 16-channel multiplexer (National Semiconductor ADC0816), a microprocessor (Hitachi HD64180), an 8-kbyte CMOS ROM (Intel 27C64), four 32-kbyte CMOS RAM's (NEC μPD43256), and interfacing I/O circuits. Fig. 1 shows its circuit diagram.

Interlink pressure sensors (Interlink Electronics, Santa Barbara, CA) of 1.1 cm diameter are used to measure plantar pressures [17]. They are 0.5 mm thick and cost about $1 each. The sensor consists of a conductive polymer film and metal lands assembled together with an O-ring spacer. The response of the sensor to pressure is logarithmic with effective pressures spanning a range of 0 to 2 MPa. Based on the full scale, the hysteresis is 8%, nonrepeatability 7%, and temperature drift −0.5%/°C [17].

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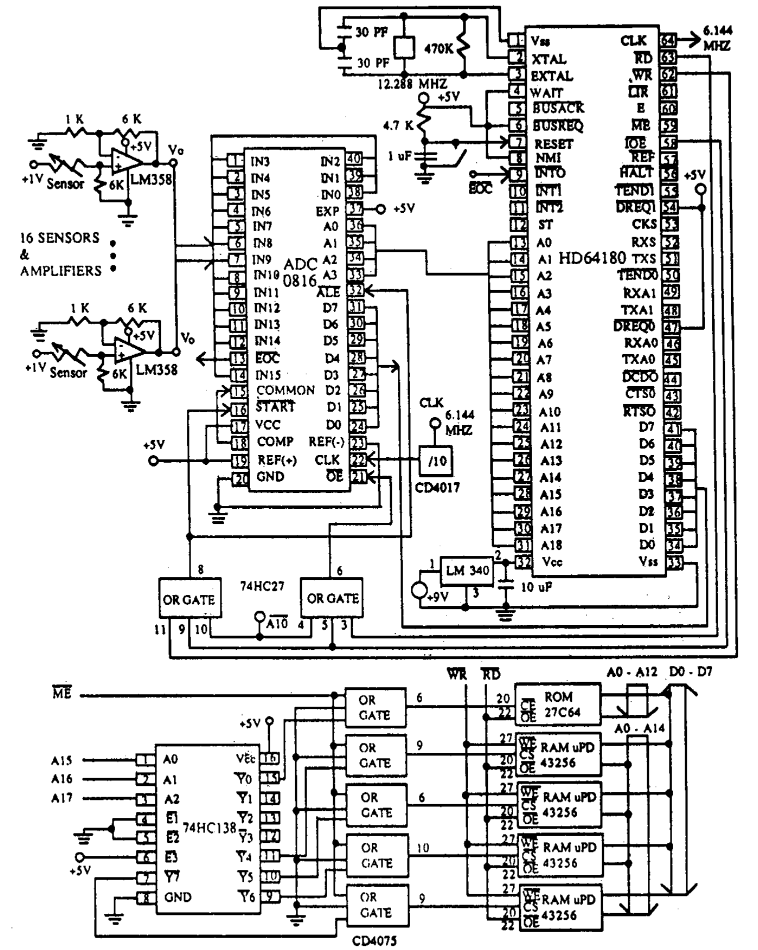

Fig. 1. Circuit diagram of the portable microprocessor-based data-acquisition system.

Sensor locations were clinically determined for each subject by recording walking patterns on an APEX footprint mat that had been evenly inked and covered with paper. Seven relatively darker (highly loaded) areas under the anterior and posterior heel, hallux, first metatarsal, second/third metatarsal, fourth metatarsal, and fifth metatarsal were selected for sensor locations. To compensate for nonlinearity and temperature drift, all sensors were dynamically calibrated at 36°C after they had been located within the insole. The outputs of the sensor and a calibration cell were used to generate piecewise linear lookup tables for translation of voltage to pressure.

The sensors in each insole are connected to the amplifier circuits through a 14-wire flat ribbon cable. Each amplifier yields an output of 0–3.5 V for the range of pressures from 0–2 MPa to meet the input requirement of the ADC. We selected LM358 amplifiers because of their low power consumption and operation on a single positive power supply. The ADC provides for sampling of 16 input channels by control of the on-chip multiplexer.

The 8-bit CMOS microprocessor provides high performance and low power consumption. It incorporates numerous on-chip system functions. On-chip features that are used included memory management and two 16-bit programmble reload timers (PRT). The microprocessor consumed 19 mW of power at 6 MHz operation by providing a SLEEP mode and a SYSTEM STOP mode. It consumed 75 mW at 6 MHz operation in the active mode. The microprocessor selects sensor channels through the ADC multiplexer and controls the sampling rate through a programmable reload timer.

The operation cycle begins with the microprocessor initiating data address and ADC channel pointers, setting the PRT registers, and enabling interruption and SLEEP. The microprocessor selects a signal channel and starts conversion. The conversion of each sampled data point requires approximately 100 μs. The end of conversion (EOC) signal interrupts the microprocessor, which then reads an 8-bit sample from the ADC and stores it in memory. The 8-bit conversion provides 20 mV of amplitude resolution for voltages ranging between 0 and 5 V which corresponds to an 11 kPa pressure resolution with a respective range of 500–700 kPa.

The sampling rate for analog-to-digital conversion was selected based on both published and experimental data. Antonsson and Mann reported that 98% of the spectral power from bare-foot walking across a Kistler force plate was below 10 Hz and over 90% below 5 Hz [18]. Acharya *et al.* reported similar findings for in-

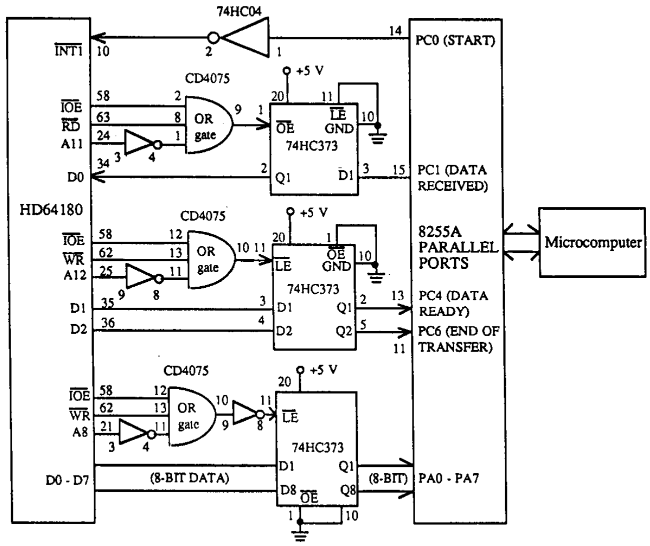

Fig. 2. Parallel interfacing circuits for data transfer between the microprocessor and the microcomputer (IBM PC).

shoe plantar pressures [19]. In order to establish an adequate sampling rate for our experiments, we sampled at different rates ranging from 5–200 samples per second (sps). Time domain comparison of the data showed that the signal sampled at 20 sps was not significantly different from that at 200 sps. Thus, 20 sps provides adequate pressure resolution and increases recording time during walking. Our system design permits adjustment of the sample rate (by hardware switch) up to 100 sps to study higher frequency pressures (utilizing the same software) such as those obtained during running or jumping activities. For extended periods of recording, the unit is programmed to sample briefly but at regular intervals. Recently, pressure data acquired for 5 s each minute extended normal ambulation tests to two hours. When the 120-kbyte data memory is full, a memory-full indicator flashes an LED at 5 Hz with a 5% duty cycle. Power for the portable unit is supplied by six 1.5-V AA alkaline primary batteries and regulated to an output of 5.0 V (LM340). The portable unit draws about 30 mA of current when in the active mode and is capable of continuous operation for approximately 10 h. The power supply and unit design provide a lightweight (0.8 kg) system which minimally encumbers gait during data collection.

At the completion of a gait test, data stored in the portable unit are downloaded to an IBM PC compatible through a parallel port. Fig. 2 depicts the parallel interfacing circuits between the microprocessor and the PC. A programmable peripheral interface chip (Intel 8255A) provides the parallel interface. Its functional configuration is programmed by the PC's system software. Operationally the chip's three 8-bit ports are set so that port A is an input port and port C is for I/O control. The four high bits of port C are input bits and four lower bits are output bits. The port B is not used. The PC first interrupts the microprocessor to initiate data transmission. The microprocessor then writes data to port A and sets the data ready line. With data-ready set, the PC reads the data on port A and stores it. The PC sends a data received signal acknowledged by the microprocessor by resetting the data ready line. If the end of transfer bit is not set, the PC continues the communication link. It takes about 30 s to transmit 120 kbytes of data to the PC. The recorded voltage data are converted into pressures with prestored sensor calibration lookup tables.

Experimental Results

To demonstrate the capability of the portable data-acquisition system, multiple walking tests of a normal adult subject (male, weight: 90 kg, and height: 180 cm) were conducted. The subject walked continuously for a period of 4 min at his natural cadence of 108 steps/min (velocity: 1.44 m/s) with the assistance of a metronome. A sample of collected pressure data is shown in Fig. 3. Computed averages of peak pressure, foot-floor contact duration, and pressure-time integral by sensor location for the 394 steps are presented in Tables I–III. A clinical study of 10 normal and three diabetic insensate subjects using the current system is described elsewhere [40], [41].

Discussion and Conclusion

Peak plantar pressures obtained with our portable unit for a normal subject are in the range of 434 to 1041 kPa, which are consistent with results reported by others [4]–[4], [44]–[44]. Bauman and Brand found peak pressures in shod walking to be within 90 to 394 kPa under the hallux, first metatarsal, second metatarsal, fifth metatarsal, and heel [4]. Soames *et al.* reported discrete plantar pressures in the range of 100–480 kPa [3]. Gross and Bunch reported peak pressures of 140–450 kPa with the highest pressures at the hallux and the lowest pressures at the fifth metatarsal [4]. Cavanagh and Michiyoshi found peak pressures from walking in tennis shoes to be 600 kPa under the hallux and second toe for a normal subject [44]. Clarke found an overall mean peak pressure for bare-foot walking of 433 kPa with a force plate [43]. In another study, Soames found overall peak pressures in the range of 150–550 kPa under the plantar surface during shod walking [44].

The portable microprocessor-based data-acquisition system described here allows natural unencumbered monitoring of foot-to-shoe contact pressures during various daily living activities. The portability and relative low cost of the unit suggest its use as a

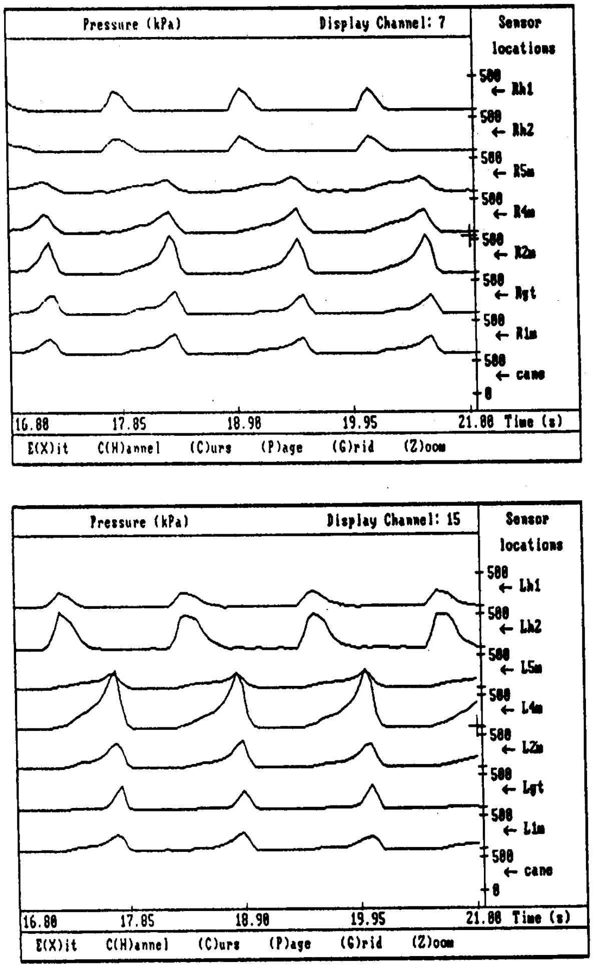

Fig. 3. Plantar pressures as a function of time during walking for a normal subject.

TABLE I
AVERAGE PEAK PRESSURES (kPa) FOR NORMAL WALKING (SD: STANDARD DEVIATION)

Locations	Right Foot (SD)	Left Foot (SD)
Posterior heel	380 (47)	264 (20)
Anterior heel	242 (26)	649 (46)
5th metatarsal	234 (29)	299 (41)
4th metatarsal	368 (38)	1041 (135)
2nd metatarsal	678 (110)	506 (52)
Great toe	322 (22)	378 (67)
1st metatarsal	328 (49)	289 (72)

TABLE II
AVERAGE FOOT CONTACT DURATION (ms) FOR NORMAL WALKING (SD: STANDARD DEVIATION)

Locations	Right Foot (SD)	Left Foot (SD)
Posterior heel	292 (43)	430 (76)
Anterior heel	356 (59)	436 (71)
5th metatarsal	681 (31)	678 (30)
4th metatarsal	601 (28)	582 (34)
2nd metatarsal	459 (60)	571 (34)
Great toe	460 (73)	250 (27)
1st metatarsal	569 (44)	556 (67)

TABLE III
AVERAGE PRESSURE-TIME INTEGRAL DATA (kPaS) FOR NORMAL WALKING (SD: STANDARD DEVIATION)

Locations	Right Foot (SD)	Left Foot (SD)
Posterior heel	55(8)	58 (9)
Anterior heel	43 (7)	148 (23)
5th metatarsal	80 (11)	85 (12)
4th metatarsal	97 (17)	234 (39)
2nd metatarsal	124 (30)	111 (16)
Great toe	56 (6)	42 (8)
1st metatarsal	73 (14)	66 (15)

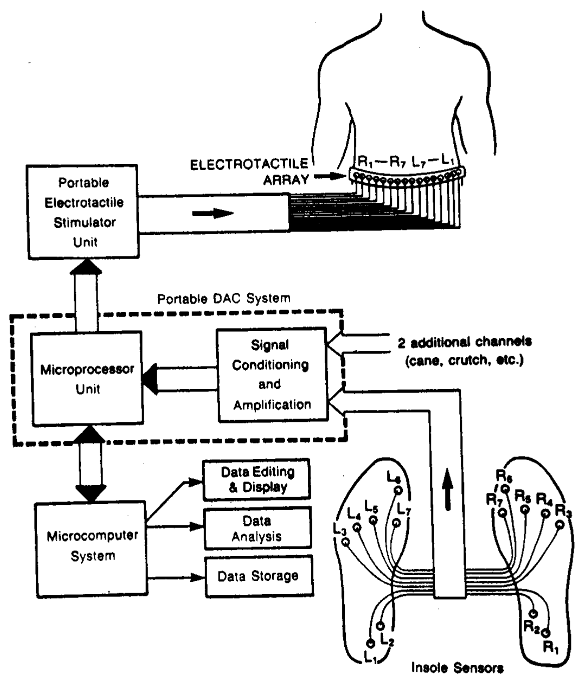

Fig. 4. Application of the portable unit for sensory substitution.

practical clinical monitoring tool. In our laboratory, the system is being used to investigate the role of plantar pressures in causing soft tissue damage in diabetic insensate feet [40] and differences in plantar pressures during walking and shuffling [41]. In this study, pressure metrics are being investigated as part of an overall sensory substitution system shown in Fig. 4. The role of the microprocessor in this system is to include not only data acquisition but control of electrotactile stimulation as well [45]. Future applications include sensory substitution for insensate feet and mobility aids for the blind.

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A Layer-Recursive Formula for the Medium Filter for Volume Conductor Problems in Radially Symmetric Layered Media

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Abstract—**Simplified algorithms for the computation of the filter coefficients used in solutions of the forward and inverse volume conductor problems in a multilayered cylindrical geometry are derived. The new algorithms are layer-recursive, as opposed to previous algorithms which were specific for the structure studied. The new algorithms not only eliminate the need to derive algebraically cumbersome filter expressions, but also speed up their numerical evaluation.**

I. Introduction

Recently, Fourier expansions [1] have been applied to volume conductor problems in multilayered cylindrical geometries. Briefly, the Fourier method entails modeling the action of a volume conductor upon evoked potentials as an equivalent filter. This allows potential field computations to be performed using signal processing operations.

In [2], a two-layer cylindrically symmetric model was used to describe the internodal region of myelinated nerve fibers. A multilayered model was also employed in [3], [4] to describe a peripheral limb containing nerve fiber bundles. In both cases a Fourier expansion was used along the cylindrical axis, and a Bessel function expansion was used in the angular direction. Although the use of the Fourier method simplifies computations of potential fields, the derivation of the filter functions for multilayered media is an algebraically tedious procedure resulting in cumbersome expressions [2], [3].

This communication describes a computationally simple layer-recursive procedure for the derivation and numerical evaluation of filter coefficients, which exploits the mathematical structure of the filter expressions. The method can be used to compute filters for arbitrary layered structures excited by either radially symmetric or eccentric (off-axis) sources. This layer-recursive form has the following advantages:

1) In [2], [3] the boundary conditions at each interface, and at infinity, were combined into a $2n \times 2n$ system of equations where n is the number of layers. Although no details were given, this system was evidently solved algebraically, leading to cumbersome expressions. In contrast, the layer-recursive procedure merely requires multiplying $n - 1$ 2×2 matrices, followed by a scalar division; this procedure effectively solves the $2n \times 2n$ system of equations.

2) The procedure is layer-recursive; not only can the number of layers be increased with minimal additional computation, but the medium filters at all locations within the medium are automatically computed as well. This is important in the radial formulation of the problem of active source location estimation [5].

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