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A Holter-type, microprocessor-based, rehabilitation instrument for acquisition and storage of plantar pressure data

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Abstract—A Holter-type, microprocessor-based, portable, in-shoe, plantar pressure data acquisition system has been developed. The system allows continuous recording of pressure data between the sole of the foot and the shoe during the performance of daily living activities. Fourteen conductive polymer sensors acquire the plantar pressure history, which is then stored in the system memory. Pressures are sampled at a rate of 40 Hz from each of the 14 sensors for up to 8 hrs. The extended recording and processing capacity of the system developed in this study allows quantitative analysis of cumulative plantar pressure and temporal gait data necessary for characterization of event-related alterations in plantar pressures. The alterations that could be examined with the system include rehabilitative, therapeutic, surgical, and nonsurgical treatment. The system is fully portable and does not disrupt the natural gait pattern of the subject during ambulation. Peak plantar pressures, pressure-time integrals, and contact durations are determined for each of the insole sensors.

Key words: *gait, Holter, microprocessor, plantar pressure, sensors.*

INTRODUCTION

There are many clinical situations in which it is useful to measure forces exerted upon the plantar surface of the foot. Force plate dynamometers have been available for several years, and more recently insole plantar pressure measurement devices have become available (1–3). A limitation of both types of systems, however, has been a capture interval restricted to seconds or minutes. At best, only a few consecutive steps may be studied, which may give a misleading picture of what forces the plantar surface of the foot experiences during the thousands of steps that occur in the span of a normal day. A system capable of continuously recording plantar pressures for an entire day has not been previously cited in the literature. The purpose of this report is to describe a Holter-type data acquisition system that provides long-term and continuous recording capability of in-shoe plantar pressures for up to 8 hrs.

METHODS

Data Acquisition System

The data acquisition system consists of 14 conductive polymer Force Sensing Resistors™ (Model 302A, Interlink Electronics, Santa Barbara, CA) sensors (FSR), a +5.0 V micropower voltage regulator (LP2940AC), 4 low-voltage micropower quad operational amplifiers (OP490GS), and a low-power 12-bit sampling analog-to-digital (A/D) converter (MAX190BCWG). An 8-bit DS5001FP-16 microprocessor (Dallas Semiconductor Microprocessor, Dallas, TX) operating at a crystal clock frequency of 11.0592 MHz is utilized with a monolithic 16-channel analog multiplexer (DG506ACWI), 32-kbyte

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high-speed CMOS static RAM (HM62256A), 32×512 kbyte, high-speed pseudo-static RAMs (658512LFP-10), 5×3-to-8 line decoders (74HC138A), a +5.0 V powered RS-232 Driver/Receiver (MAX232CWE), and other interfacing I/O circuitry. Visual output is provided by a 20 character × 4 line dot matrix liquid crystal display module (OPTREX, DMC20434). The system circuitry is assembled on a double-sided printed circuit board and employs surface mount technology (SMT). The system is powered by five 1.2 V, C-size, Ni-Cd rechargeable batteries. The unit draws a maximum of 50 mA of current, and is capable of continuous "power on" operation for up to 36 hrs before the batteries need recharging. The system is mounted in a plastic case (15×20×10 cm) and weighs 1.25 kg. The system architecture is illustrated in **Figure 1.**

The system is fully portable; subjects carry it in a belt pack and ambulate freely without disruption to their usual gait pattern (**Figure 2.**) It allows real-time recording of both pressure and gait parameters for up to 8 hrs (dependent on sampling rate) during normal daily activities. Both continuous and intermittent modes of operation are available to further expand the run time. The system is equipped with a liquid crystal display (LCD) module that indicates the current status of the data acquisition, the battery voltage level, the number of bytes recorded, and other operational messages.

The microprocessor employs custom software that controls data acquisition and sensor calibration. The software is written in C-language and compatible with the industry standard 8051 microprocessor series (C51 Compiler, Franklin Software, Inc., San Jose, CA).

Recorded plantar pressure data and calibration data are uploaded into a personal computer (PC) for further processing, analysis, and display. Command communication between the portable unit and the PC is handled through the PC serial port at a baud rate of 9,600 bits/s. Data transfer is directed through the PC parallel printer port at an approximate rate of 8,000 bytes/s. Additional PC software is written to convert raw voltage data into pressure metrics, determine various gait parameters, conduct statistical analysis, and display analysis results.

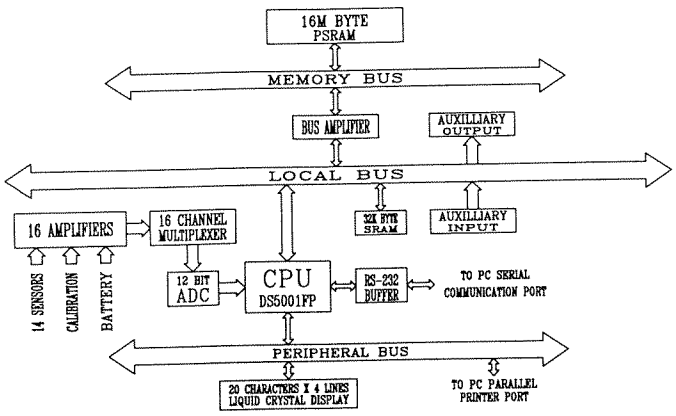

Figure 1.
The Holter-type, plantar pressure, data acquisition system architecture.

Pressure Sensors

FSRs, commonly termed conductive polymer force (pressure) sensors, were selected to measure in-shoe plantar pressures (4–7). The FSR offers numerous advantages, such as flexibility, durability, reliability, overload tolerance, electronic simplicity, and low cost (less than $4.00 each), over other types of available sensors.

In 1988, Maalej et al. explored the static and dynamic characteristics of FSRs and reported that the hysteresis

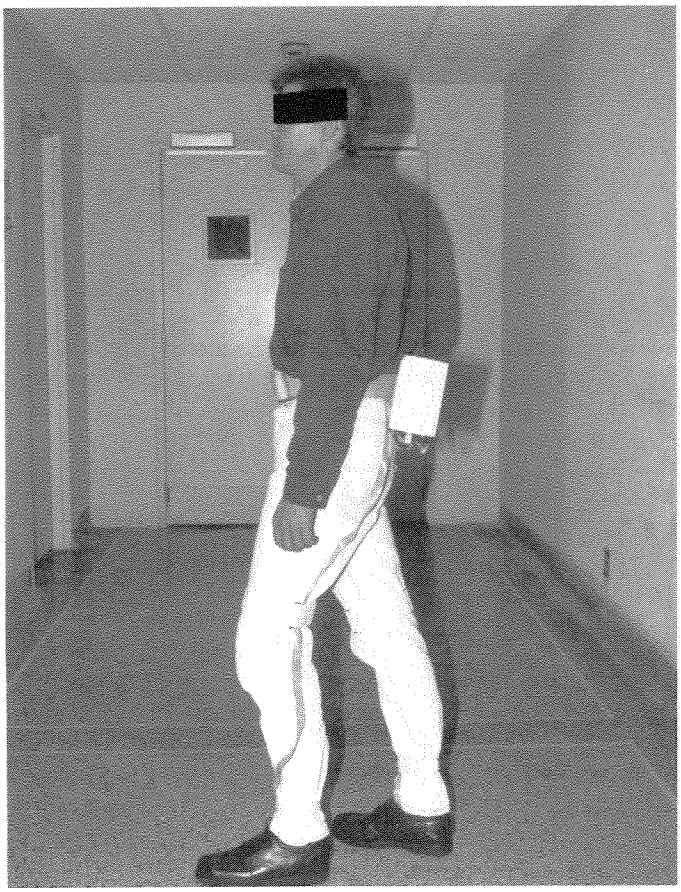

Figure 2.
A nonimpaired subject instrumented with the data acquisition insole system.

was between 5 and 10 percent of full pressure scale of 0 to 1.2 MPa (6). The nonlinearity and hysteresis characteristics of the sensor were compensated by using calibration lookup tables. Since the sensor is sealed, it is insensitive to humidity (1). The Model 302A FSR (18-mm outer diameter, 15-mm active sensing diameter, 0.4-mm overall thickness) was used in this study.

Sensor Placement

Sensors were located at seven predetermined plantar points of each foot under the calcaneus, metatarsal heads, and great toe. These locations have been shown to present pressure gradients closely associated with biomechanical variations and foot pathology in prior studies (8–13). Sensor locations for each individual were identified after investigating numerous clinical footprint techniques designed to reveal precise areas of high pressures under bony prominences of the foot (1,2). These included the APEX foot imprinter (APEX, S. Hackensack, NJ), microcapsule socks, Fuji Pressure Mat, and Shutrack system. The APEX foot imprinter was found to work best for applications in terms of accuracy, simplicity, and cost (1,2), and was selected in this study to determine sensor locations within the insole. Each subject walked barefoot on an APEX footprint mat that had been evenly inked and covered with APEX orthotic paper. The test was repeated three times; subsequently, the locations of the highest pressure area centers were averaged and determined. The operator then aligned the APEX paper on the insole to lay out the primary sensor locations as indicated by the density of ink on the impression. Upon completion, the same test was conducted to identify sensor locations on the contralateral foot. An APEX foot imprint from a nonimpaired individual is shown in **Figure 3.**

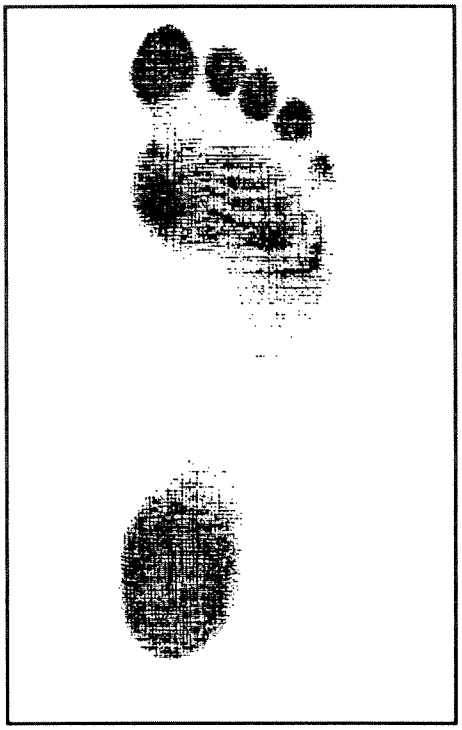

Figure 3.
An APEX foot imprint from a nonimpaired individual.

Insole Instrumentation

To reduce inaccuracies due to sensor bending, a small stainless steel disc was mounted to the back of each sensor to keep it flat. Sensors were mounted into the sculpted insole material. The upper insole layer was carefully carved to accept the sensors and thin metal backings, which were flush-mounted with the upper insole surface. Sensor cables were embedded within narrow channels carved on the upper insole material. Subjects could not perceive the presence of the sensors or cables in the insole while ambulating. The instrumented insoles were fitted into a pair of Extra Depth Easy Sport athletic shoes (P.W. Minor, Batavia, NY). Acclimation to the experimental shoes and establishment of a constant temperature shoe environment was provided during a 30-min pretest stabilization period. **Figure 4** illustrates an instrumented insole with seven discrete pressure sensors. Sensors were located at the anterior heel (AH), posterior heel (PH), first metatarsal head (1M), second metatarsal head (2M), fourth metatarsal head (4M), fifth metatarsal head (5M), and hallux or great toe (GT).

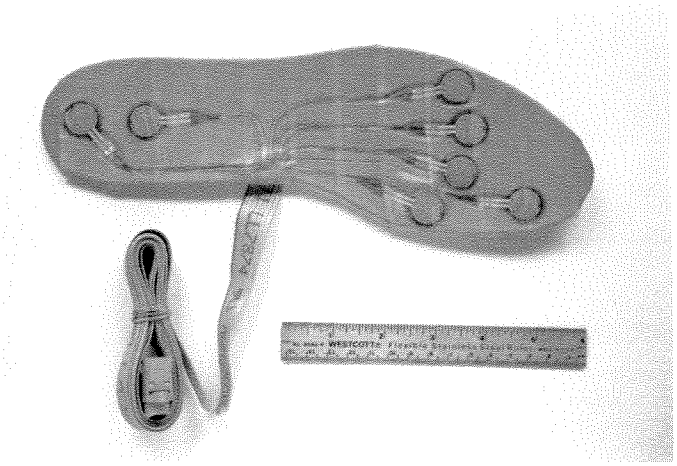

Figure 4.
An instrumented insole with seven discrete pressure sensors.

Dynamic Sensor Calibration

To better characterize sensor output during the stance phase of walking, a dynamic force application unit consisting of a compression lever, a precalibrated 440 N strain gage load cell, and preamplifier was used to dynamically calibrate the FSRs (**Figure 5**). Dynamic loads with durations similar to those of stance phase foot contact (≤620 ms) were applied to each sensor during calibration. Loads were assumed to be distributed over the entire sensing area of the FSR. Resulting calibration data from each sensor and the load cell were then automatically transferred to the PC, where a piecewise linear calibration table converted voltage data into pressure values. Pressure computations were determined by dividing the measured forces over the assumed contact area. A typical dynamic calibration curve for an FSR is presented in **Figure 6.** To compensate for temperature sensitivity, a Plexiglas™ oven, surrounding the calibration apparatus and the instrumented insole, was used to calibrate the sensors at 36 °C. This temperature simulated the in-shoe temperature environment. The oven consisted of two 100 W light bulbs, a small air circulating fan, a temperature sensor, and a temperature regulator device (1).

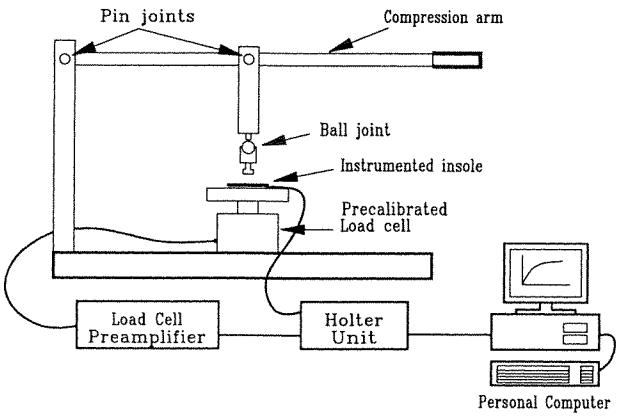

Figure 5.
Schematic representation of the dynamic force application unit.

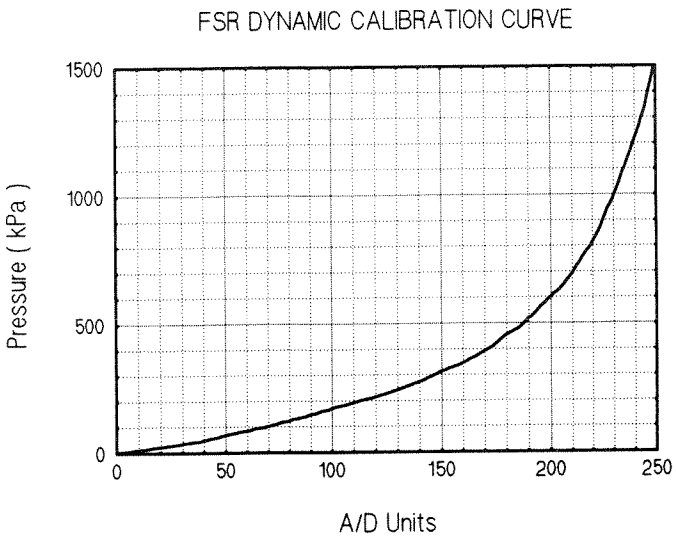

Figure 6.
A typical dynamic calibration curve for one of the Force Sensing Resistors™.

Signal Conditioning and Digitization

Signals from the FSRs used to sense insole pressures are processed by the four amplifiers, which double the input signal magnitude. Output signals from the amplifiers range from 0 to 4.095 V and correspond to pressure values ranging from 0 to 1.2 MPa. The amplified signals are then switched to the A/D converter by the multiplexer. The resulting digital data are collected by the microprocessor and stored in the appropriate memory location.

The signal-to-noise ratio (S/N) was determined for the two types of signals associated with the system. These are signals from the FSRs and the strain gage load cell preamplifier (used for sensor calibration). The S/N was calculated at different steady load values, ranging between no load and full scale load. The FSR input presented a signal-to-noise ratio of 64.6 dB, while the strain gage load cell presented a signal-to-noise ratio of 51.3 dB. The S/N value obtained for the FSR input reflects the simple passive nature of the sensor, while the value obtained for the strain gage load cell signal reflects the effect of the preconditioning circuitry.

During data acquisition, the signals are sampled at a 12-bit resolution (1/4096), and subsequently stored with 8-bit resolution (1/256) by truncating the lower 4 bits. Mathematically, this is equivalent to an "integer divide" by 16 (4096/256). The peak noise level observed in the FSR signal was ±1 count in 4096 levels, while the peak noise level observed in the strain gage load cell signal was ±7 counts in 4096 levels. When dividing by 16, these values (±1 and ±7 counts) will result in a value less than ±1 part in 256 (0.4 percent). Based on the above results, no additional filtering was required.

Plantar pressure data were sampled at a rate of 40 samples per second (sps), a rate that met the Nyquist criterion for data acquisition (2,14,15). Antonsson and Mann reported that 98 percent of the spectral power from

barefoot walking across a Kistler force plate was below 10 Hz and over 90 percent was below 5 Hz (15). In 1989, Acharya et al. reported similar findings for in-shoe plantar pressures (14). In 1991, Zhu et al. conducted time domain analyses to establish an adequate sampling rate for in-shoe plantar pressure measurements (2). The authors investigated a series of sampling frequencies ranging from 5 Hz to 200 Hz and reported that signals sampled at 20 Hz were not significantly different from those sampled at 200 Hz. Although the results obtained from these studies were based on nonpathologic populations, the energy bounds were extrapolated in our clinical studies of pathological and nonpathological cases.

System Characterization

The Holter-type, microprocessor-based, portable, in-shoe, plantar pressure data acquisition system was characterized using multiple tests. The system yielded a voltage range from 0 to 4.095 V with a resolution of 1 mV. The entire 16-Mbyte storage capacity of the system was verified during 8 hrs of continuous recording from 14 channels at a sampling rate of 40 Hz. Software was used to verify accuracy in data storage and data transfer through the serial and parallel ports by transferring known data patterns from the Holter unit to the PC (16). The transfer rate through the parallel printer port was approximately 8,000 bytes/s, requiring less than 35 minutes for 16 Mbytes of data to be uploaded. The LCD module was verified for full screen (20 character × 4 line) display. A summary of the Holter-type data acquisition system specifications is illustrated in **Table 1.** This phase of characterization was simply designed to demonstrate feasibility. Reliability and analyses of time-related pressure alterations, which may occur during daily living activities remain as future project goals.

Table 1.
The Holter-type, plantar pressure data acquisition system specifications.

Maximum Channels	16 channels
Data Memory (PSRAM)	16,711,683 bytes
Program and Data Memory	32,768 bytes
Maximum Sample Rate	40 samples per second
Maximum Recording Time	8 hours (@ 40 Hz)
Resolution	12 bits
Dot Matrix LCD Module	20 character x 4 lines
PC Interface	1 serial port, 1 printer port
Command Data Rate	9600 baud
Data Transfer Rate	8,000 bytes/second
Power Draw	50 milliamperes
Battery	6 volt, 36 hours
Size	15 x 20 x 10 cm
Weight	1250 grams

RESULTS AND DISCUSSION

The portable system was used to acquire plantar pressure data from an adult male subject during the activities of a usual daily working environment (6–8 hrs of continuous recording). The subject (age: 38 years, height: 181 cm, weight: 73 kg, and shoe size: U.S. men's 9 1/2) was free from any orthopaedic, neurological, or systemic complaint and did not suffer from foot disorders or other abnormalities. He was evaluated with the APEX footprint mat for definition of high load areas under each foot surface, and a customized pair of insoles was crafted with seven sensors for each foot.

Multiple walking tasks of the subject were monitored during 3 working days, spaced 1 week apart. The subject was asked to carry the Holter unit in a belt pack and ambulate freely about the working environment at his freely selected cadence. He could not perceive the presence of the sensors in the insole while ambulating and did not report discomfort or fatigue during the test trials. The subject was instructed to keep a logbook of activities to record observations regarding the system acceptability or any problems encountered. A technical record of system performance was also maintained in the laboratory regarding sensor performance, cables and connections, insole, and hardware performance. Plantar pressure data were recorded during various tasks, such as sitting at a desk with no plantar activity, level walking, stair climbing, stair descent, and standing still. The system was capable of demonstrating changes in plantar pressure patterns during these various tasks. A sample of collected pressure data at the first metatarsophalangeal joint (1M) is depicted in **Figure 7.**

Peak plantar pressures, pressure-time integrals, and contact durations were determined for each of the insole sensors. These metrics have proven useful in prior clinical studies (1–3,8–13,17,18). In a classic work, Kosiak demonstrated that for ischemic ulceration, there is an inverse relationship between pressure and time (19). Since then, it has become well accepted that the development of ulceration is not only related to the amount of exerted pressure but also the amount of time (duration) the pressure is applied. The pressure-time integral is a combined

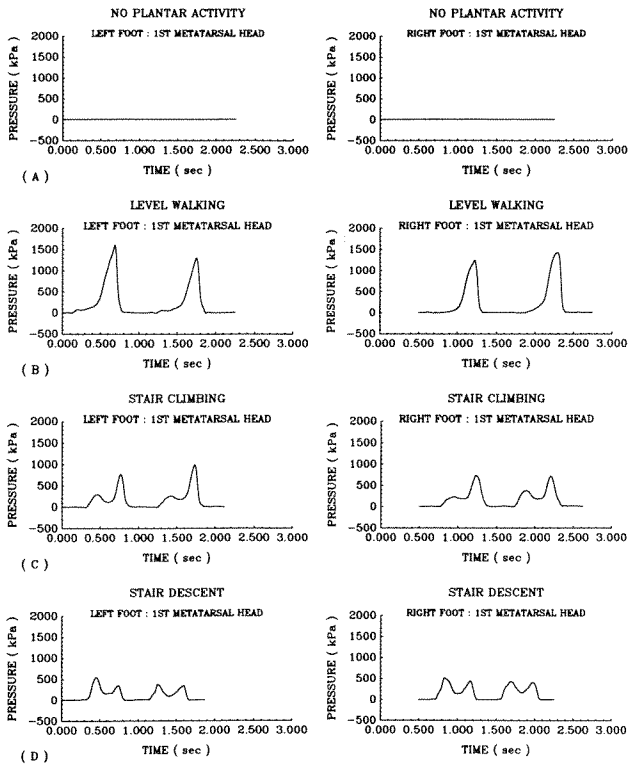

Figure 7.
A sample of collected pressure data at the first metatarsal head during: A) no plantar activity; B) level walking; C) stair climbing; and D) stair descent.

metric, as it examines not only the peak pressures, but also the amount of time (duration) the site is loaded.

During a continuous level walking test, the subject freely selected a cadence of 107 steps/min on a 100 m concrete walkway. Steps around the ends of the walkway were excluded to eliminate any altered gait patterns during the turn maneuver. The walking speed was 1.21 m/sec, the stride length was 1.36 m, and the total number of steps was 2,944. Results for the test trial are illustrated in **Table 2.** Mean, standard deviation, and coefficient of variation values for peak pressures (kPa), contact durations (msec), and pressure-time integrals (kPa•sec) are provided in the table for each of the insole sensors. Mean peak pressures ranged from 158.7 kPa to 674.9 kPa, while mean contact durations ranged from 279.6 msec to 600.3 msec, and mean pressure-time integrals ranged from 28.2 kPa•sec to 139.6 kPa•sec. These results compare favorably with those reported by others (3,17,18). In 1963, Bauman and Brand found peak pressures during shod walking to be within 90 and 392 kPa (3). In 1982, Soames et al. reported on discrete peak pressures in the range of 100 to 480 kPa (18). In 1988, Gross et al. measured peak pressures in the range of 120 to 450 kPa (17). The mean and standard deviation values for the test measurements are depicted in **Figure 8.**

Table 2.
Foot plantar pressure distribution of the test subject during level walking (n=2,944 steps).

	Peak Pressure (kPa)			Contact Duration (msec)			Pressure-Time Integral (kPa•sec)		
Sensor Location	**Mean**	**S.D.**	**Coef. Var.**	**Mean**	**S.D.**	**Coef. Var.**	**Mean**	**S.D.**	**Coef. Var.**
LEFT FOOT									
PH	481.6	45.3	0.094	314.8	46.1	0.147	88.1	12.8	0.146
AH	461.3	61.0	0.132	384.0	69.3	0.180	99.5	22.0	0.221
1M	628.6	122.3	0.195	326.1	36.9	0.113	98.6	21.5	0.218
2M	306.9	47.8	0.156	339.5	47.9	0.141	53.6	7.1	0.132
4M	410.3	61.3	0.149	462.5	67.9	0.147	88.1	16.5	0.187
5M	266.9	72.3	0.271	461.7	108.2	0.234	74.2	27.1	0.365
GT	179.5	38.8	0.216	279.6	26.2	0.094	28.2	6.3	0.225
RIGHT FOOT									
PH	318.8	36.3	0.114	299.8	38.7	0.129	53.4	9.1	0.170
AH	370.3	59.7	0.161	346.4	52.8	0.152	72.9	16.2	0.222
1M	674.9	125.5	0.186	304.8	35.5	0.116	115.3	23.8	0.207
2M	644.5	105.8	0.164	458.5	67.3	0.147	139.6	20.9	0.150
4M	575.2	105.9	0.184	497.0	41.2	0.083	137.0	23.1	0.169
5M	158.7	50.7	0.320	600.3	26.0	0.043	51.2	13.2	0.257
GT	337.6	69.1	0.205	307.3	49.5	0.161	49.3	8.6	0.174

S.D. = standard deviation; Coef. Var. = coefficient of variation values.

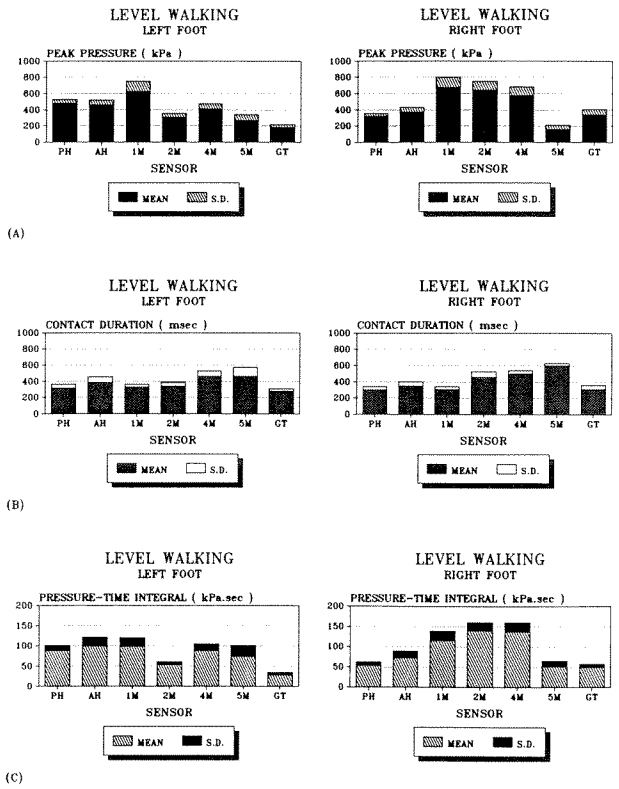

Figure 8.
Mean and standard deviation values by sensor during level walking, (n=2,944 steps): A) peak pressures; B) contact durations; and C) pressure-time integrals.

The sensors, with their thin metal backings, offered proportionately greater stiffness than the other components of the insole structure. Previous tests by our group indicated that sensor/backing stiffness exceeded that of the insole by at least 20:1 ratio. A major effect of the compliance difference is insole distortion about the sensor edges. The effect of this distortion on adjacent sensors remains as a topic for future investigation, in which the foot, insole, and sensor structures are possibly modeled with the finite element method. On the basis of sensor and system calibration and characterization studies, the overall system error was estimated to be between 7 and 14 percent. The primary sources of error were sensor hysteresis (5–10 percent error), sensor analog drift (1–2 percent error), and system electronics drift (1–2 percent error).

The results obtained from this study indicate that the system is appropriate for further clinical application and for characterization of event-related alterations including rehabilitative, therapeutic, surgical, and nonsurgical treatment. In our lab, the system is now being used to investigate plantar pressure distribution characteristics during ambulation in nonimpaired adults and those with foot and ankle pathology.

CONCLUSION

A Holter-type device capable of long-term and continuous recording (up to 8 hrs at a sampling rate of 40 Hz) of in-shoe plantar pressures was designed and developed using existing and proven technologies. The foot pressure sensor was selected so as not to interfere with natural gait patterns. Accordingly, the sensor exhibited a thin profile so that its presence could not be perceived by the subject. The chosen sensor was durable yet capable of withstanding repetitive gait cycles with high sensitivity and the capability of withstanding large overloads. The sensor presented a short time response and low power consumption. The sensor also was wear resistant with low hysteresis. Moderate nonlinearity was acceptable, since it could be compensated in the data processing software. The sensor also presented a well-characterized sensitivity to temperature and humidity. The FSR was able to measure pressures in the range of 0 to 1.2 MPa with high stability and repeatability.

The data acquisition system was designed to be fully portable so that the gait being studied was not constrained by a tether in a limited laboratory environment. Sufficient memory was supplied to store the complete pressure-time data at all sensor sites. The data sampling rate was adjustable and the system was able to run for long periods without a battery change. A display screen was used to indicate the current status of the data acquisition system, the battery voltage level, the number of bytes recorded, and other contingency messages.

REFERENCES

1. Wertsch JJ, Webster JG, Tompkins WJ. A portable insole plantar pressure measurement system. J Rehabil Res Dev 1992:29(1):13–8.
2. Zhu H, Harris GF, Wertsch JJ, Tompkins WJ, Webster JG. A microprocessor-based data-acquisition system for measuring plantar pressures from ambulatory subjects. IEEE Trans Biomed Eng 1991:38(7):710–4.
3. Bauman JH, Brand PW. Measurement of pressure between foot and shoe. Lancet 1963:629–32.
4. Patel A, Kothari M, Webster JG, Tompkins WJ, Wertsch JJ. A capacitance pressure sensor using a phase-locked loop. J Rehabil Res Dev 1989:26(2):55–62.

5. Bhat S, Webster JG, Tompkins WJ, Wertsch JJ. Piezoelectric sensor for foot pressure measurements. In: Proceedings of the 11th Annual International Conference of the IEEE Engineering in Medicine & Biology Society, 1989:1435–6.
6. Maalej N, Bhat S, Zhu H, et al. A conductive polymer pressure sensor. In: Proceedings of the 10th Annual International Conference of the IEEE Engineering in Medicine & Biology Society, 1988:770–1.
7. Kothari M, Webster JG, Tompkins WJ, Wertsch JJ, Bach-y-Rita P. Capacitive sensors for measuring the pressure between the foot and shoe. In: Proceedings of the 10th Annual International Conference of the IEEE Engineering in Medicine & Biology Society, 1988:805–6.
8. Wertsch JJ, Loftsgaarden JD, Harris GF, Zhu H, Harris JL. Plantar pressures with contralateral versus ipsilateral cane use (Abstract). Arch Phys Med Rehabil 1990:71:772.
9. Zhu H, Wertsch JJ, Harris GF, Loftsgaarden JD, Price MB. Foot pressure distribution during walking and shuffling. Arch Phys Med Rehabil 1991:72:390–7.
10. Zhu H, Wertsch JJ, Harris GF, Alba HM. Effect of walking cadence on plantar pressures. Arch Phys Med Rehabil 1995:76:1000–5.
11. Zhu H, Wertsch JJ, Harris GF. Asymmetry of plantar pressure during normal walking (Abstract). Arch Phys Med Rehabil 1990:71:808.
12. Zhu H, Wertsch JJ, Harris GF, Alba HM, Price MB. Sensate and insensate in-shoe plantar pressures. Arch Phys Med Rehabil 1993:74:1362–8.
13. Wertsch JJ, Frank LW, Zhu H, Price MB, Harris GF, Alba HM. Plantar pressures with total contact casting. J Rehabil Res Dev 1995:32(3):205–9.
14. Acharya KR, Harris GF, Riedel SA, Kazarian L. Foot magnitude and spectral frequency content of heel strike during gait. In: Proceedings of the 11th Annual International Conference of the IEEE Engineering in Medicine & Biology Society, 1989:826–7.
15. Antonsson EK, Mann RW. The frequency content of gait. J Biomech 1985:18:39–47.
16. Vengsarkar AS, Abler JH, Abu-Faraj ZU, Harris GF, Wertsch JJ. Holter system development for recording plantar pressures: software development. In: Proceedings of the 16th Annual International Conference of the IEEE Engineering in Medicine & Biology Society, 1994:936–7.
17. Gross TS, Bunch RP. Measurement of discrete vertical in-shoe stress with piezoelectric transducers. J Biomed Eng 1988:10:261–5.
18. Henning EM, Cavanagh PR, Albert HT, Macmillan NH. A piezoelectric method of measuring the vertical contact stress beneath the human foot. J Biomed Eng 1982:4:213–22.
19. Kosiak M. Etiology and pathology of ischemic ulcers. Arch Phys Med Rehabil 1959:62–9.

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