Microprocessor-based plantar pressure measurement system for rehabilitation of lower limb

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Abstract

Evaluating plantar pressure plays a crucial role in the treatment of patients undergoing rehabilitation for fractures of the lower limb. During the rehabilitation process, the repeated application of a higher than usual plantar pressure is a major factor in developing formation of callus. However, it is important that an appropriate pressure loading on the basis of the patient’s weight is applied to ensure an efficient callus recovery. Consequently, this paper develops a microprocessor-based, portable, ambulatory foot pressure measurement system. In the proposed system, five pressure sensors are placed in appropriate positions on the outsole of the patient's shoe, i.e. under the medial of the heel, and at the metatarsal heads and the hallux, to measure the pressure acting on the foot. The measurement system has two important features: (1) an acoustic alarm which sounds when the weight-bearing exceeds the predetermined plantar pressure; (2) an embedded memory storage device to store the pressure data for subsequent clinical analysis during the rehabilitation process. The effective operation of the proposed measurement system is verified experimentally by its accuracy and dynamical response.

Key words: Foot pressure, Pressure sensor, Rehabilitation, Microprocessor

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1. Introduction

Physiotherapists counsel patients with lower limb fractures to use moderate exercise to aid in the healing process. Care must be made in managing the healing process as underloading the limb may slow the healing process and overloading the limb may risk refracture. For the therapist calculating maximum limb loads is difficult and even trained patients can have inconsistent results. Our approach is to provide the patient with relevant gait performance feedback during training. Several critical points in the standard gait are monitored. A discussion of the standard gait cycle, gait analysis and how the gait data is fed back to the patient follows.

The gait cycle [1-3] is defined as the period from the moment at which the heel of one foot first contacts the ground to the moment at which the same heel makes contact with the ground once again. The conventional gait cycle has two distinct phases: the stance phase and the swing phase.

(A) Stance phase: During this phase, the foot makes contact with the ground and must withstand the counterforce exerted by the ground in order to maintain the body in a stable upright position. This phase accounts for approximately 60% of the total gait cycle. This phase includes the following sub-phases:

a. Heel strike: The individual commences the first step and the heel makes progressive contact with the ground
b. Foot flat: The thenar comes into contact with the ground. This phase accounts for the first 10% of the total gait cycle.

c. Mid-stance: The thenar makes full contact with the ground and supports most of the body weight. This phase accounts for the 10%-30% stage of the gait cycle.
d. Terminal-stance: The heel of the foot starts to lose contact with the ground as the heel of the other foot first starts to make contact with the ground. This phase accounts for the 30%-50% stage of the gait cycle.
e. Push off: The front of the thenar pushes off the ground and the body’s center of mass moves forward. This phase accounts for the 50%-60% stage of the gait cycle.
f. Toe off: The toes of the foot leave the ground, signifying the beginning of the swing phase.

(B) Swing phase: The foot leaves the ground and swings through the air. This phase accounts for approximately 40% of the total gait cycle.

a. Initial swing: The foot starts to swing forward. This phase accounts for the 60%-75% stage of the gait cycle.
b. Mid-swing: The foot accelerates and moves forward, crossing the center of gravity of the body. This event accounts for the 75%-90% stage of the gait cycle.
c. Terminal swing: The foot begins to decelerate until the heel starts to make contact with the ground once again. This event accounts for the 90%-100% stage of the gait cycle.

At the time that the heel of one foot first starts to come into contact with the ground, the toes of the other foot are just starting to break contact with the ground. In other words, the two feet are on
the ground at the same time, providing a so-called “double support”. This double support condition is maintained for approximately 20% of the cycle. During the remainder of the cycle, a condition of “single support” is maintained.

Performing a gait analysis when the patient is walking provides valuable information for patient training and recovery appraisal. Many researchers have conducted advanced gait analysis for the patient with artificial joints [4-9]. However, from a clinical practice perspective, it is more meaningful to conduct research aimed at understanding and controlling lower limb loading when using an assistive device. Therefore, several researchers [10-12] investigated patients’ capabilities when using a stick to compensate for an injured limb and examined the loading when walking with an extractor. A systematic appraisal approach for a patient’s walking characteristics when using an extractor was developed [10]. A mathematical method was used to calculate the patient’s loading during the stick swing phase when walking using a stick was proposed [11]. Finally, a kinetic analysis method was developed to calculate the torque and the power on the elbow and hand when walking using a stick [12]. Although the measurement data provided by these various methods enable a therapist to evaluate a patient’s loading, the appraisals can only be performed in a laboratory equipped with advanced facilities.

The studies [13-19] installed pressure converters in the different area of foot to measure the stress value of its corresponding places. Wearing et al [15] discovered that the stress distribution of an individual with a fractured limb is significantly different from that of a non-afflicted person. However, these stress distributions were obtained with the foot in an unshod status, and hence the data are not truly representative of the variations in the patient’s foot stress during the course of daily life. Lawless et al. [16] discovered that when the patient with shoe the stress distribution of foot will be averaged, and thereby avoiding over-concentration of the loading on foot. Barnett [17] developed a pedar system to measure the transient response parameters of a patient’s gait. Sung et al. [18, 19] derived a feedback control system for lower limb loading. Generally speaking, foot force measurement is helpful in analyzing the changes in loading capability of a patient’s lower limbs. The measured data provides a therapist with valuable information with which to devise an appropriate recovery program.

Drawing on the contributions of the studies presented above, this study develops a microprocessor-based, portable, ambulatory foot pressure measurement system equipped with two key features: (1) an acoustic alarm to provide a warning when the weight-bearing is above the maximum or below the minimum loading thresholds programmed specifically by each patient according to therapists’ suggestion, and (2) a flash memory-based storage device to store the pressure data for clinical analysis during the rehabilitation process. The stored data can then be used by the physiotherapist in a treatment session / evaluation session at a later date. Therefore this device enables the recovery process to be carried out in the most
efficient and least intrusive manner possible.

2. Method

This study commenced by using the procedure outlined in [20-22] to perform a gait analysis on a non-afflicted individual’s walking action. The results of this study indicated five areas of the sole which were subjected to particularly high weight loadings. Pressure sensors embedded in a proprietary outsole were then positioned in these five locations and used to measure the foot stress value of an individual with a fractured limb.

2.1 Subjects and Experimental Protocol

In order to compare the usability of the proposed system, the ground reaction forces (GRF) was measured by using two 1kHz force plates (Type 9281B, Kistler Instrument Corp., Winterthur, Switzerland) system. Two subjects (1 male, 1 female) volunteered to participate in the study. These two normal, healthy volunteers weight were 92kg and 47kg, respectively. They were asked walked at a self-selected comfortable walking speed for a trial.

2.2 Principles of force measurement

The total stress acting on the foot can be divided into a number of areas of constant pressure, \( a_i(t) \), and a sensor module can be installed in each area to measure the pressure, \( p_i \, (kg/cm^2) \). The total loading \( F \, (kgf) \) acting on the foot can therefore be calculated from:

\[
F(t) = \sum_{i=1}^{5} a_i(t) p_i \quad (1)
\]

and

\[
J(t) = \int_{0}^{t} F(s) ds \quad (2)
\]

where \( J \) is the momentum of motion.

For a patient undergoing a recovery course using the proposed pressure measurement system, a default maximum loading, \( F_{\text{max}} \), and a default minimum loading, \( F_{\text{min}} \), can be specified to each patient. If the calculated loading, \( F \), is below \( F_{\text{min}} \), the alarm emits two short “beeps”, telling the patient to increase the loading as he/she walks. If \( F \) lies between \( F_{\text{min}} \) and \( F_{\text{max}} \), the alarm will not sound, and the patient can continue walking with the same loading level. However, if \( F \) is above \( F_{\text{max}} \), the alarm will emit a continuous “beep”, alerting the patient that he/she is overloading his/her lower limb. When \( J \) exceeds the default rehabilitation momentum value, the patient needs to stop walking.

3. Measurement system design

3.1 System overview

Figure 1 illustrates the structure of the proposed measurement system. The system consists of a microprocessor chip (PIC16F877), five individual pressure sensors with associated amplifiers, a loudspeaker, a flash ROM storage device, a battery power supply, and a PC. The pressure sensors and amplifiers are integrated into five small modules and installed on the outsole. When a signal is obtained from the pressure modules (i.e. the patient starts to walk), the PIC16F877 chip transforms it into a digital signal via an internal A/D converter and then transmits the signal to the ALU to calculate the total force acting on the foot. The force value is saved and compared with the minimum and maximum
permissible force settings specified for that patient, as discussed in Section 2.2. The photograph of hardware implementation of the proposed system is shown in Figure 2.

3.2 Circuit Design

3.2.1 PIC 16F877 Microprocessor

The PIC 16F87X series chipset is an 8-bit single chip RISC microcomputer produced by Microchip Inc operating at a clock frequency of 16 MHz. This chipset includes an 8K flash memory, five I/O Ports, and supports 14 interrupts. Moreover, it supports an 8-channel analog to digital converter (A/D converter) with 10-bit resolution, Pulse Width Modulation (PWM), and a Universal Asynchronous Receiver/Transmitter (UART), etc.

3.2.2 System Power

The pressure measurement system is powered by eight 300 mA / 1.5 V batteries. Among different brands of voltage regular ICs, the 78XX (Positive voltage regulator) and 79XX (Negative voltage regulator) series are the most commonly applied. These regulators are widely used in electrical and aviation instrumentation. This study provides the 5V required for the PIC16F877 is using an IC7805, the -5V required for the instrument amplifier using an IC7905, and the 2V required for the pressure sensor using an LM317.

3.2.3 Pressure Sensors

A previous study [23] showed that the pressure on the foot is distributed as follows: 60.5% at the heel, 7.8% at the medial of the foot, 28.1% at the front of the foot, and 3.6% at the toes. Therefore, five pressure sensors are placed in the medial of the heel, and at the metatarsal heads and the hallux, to measure the pressure acting on the foot. Furthermore, the pressure sensors had to be thin and small so as not to alter the patient’s natural gait and not to injure his/her foot. In this study, we use a PS-5KA-H pressure sensor (Kyowa, Japan) to measure the plantar pressure. The physical characteristics of the sensor are summarized in Table 1. These five sensor modules taped under the foot sole are shown in Figure 3.

3.2.4 Amplifier Design

The load sensor converts the plantar pressure into a voltage. However, the magnitude of the voltage signal is very small and must therefore be amplified. This study uses AD620 amplifiers, which are low cost and low power consumption devices suitable for pressure sensing applications. Figure 4 shows the arrangement of the AD620 amplifier circuit, comprising 3 separate amplifiers. The resistance of each amplifier is 10KΩ. The overall gain of the amplifier is given by:

\[ Gain = \frac{49.4k\Omega}{R_G} + 1. \]  

Clearly, the gain (1~1000) can be adjusted by regulating the value of \( R_G \). In this study, the value of \( R_G \) is specified as 51Ω, giving a gain of 970.

3.2.5 Storage Device

This study stores the pressure measurement data using a 24LC64 8 x 8K (64K bit) Serial Electrically Erasable PROM (EEPROM) produced by Microchip U.S.A. The memory is both read and written using an Inter-Integrated Circuit (I²C). The PIC16F877 microprocessor with an I²C peripheral serves as the EEPROM controller and can save 8,000 data rapidly. These stored data in the flash
can be then processed and analyzed for clinical information.

3.2.6 Software Flowchart

Figure 5 presents a flowchart of the system software. When the patient commences the recovery session, a foot stress will occur and the sensors produce a corresponding voltage signal, which is subsequently amplified and transferred to the microprocessor to calculate the $F$. As discussed previously, if the microprocessor determines that $F$ is below $F_{\text{min}}$, the alarm will emit two short beeps. If the value of $F$ falls between $F_{\text{min}}$ and $F_{\text{max}}$, the alarm remains silent. If $F$ is above $F_{\text{max}}$, the alarm will emit a continuous beep to warn the patient that the loading is excessive.

The momentum $J$ is derived by multiplying the value of $F$ with the time of the recovery session. The computed value of $J$ is compared with the default momentum value, i.e. $J_{\text{present}}$ specified to each patient. If the value of $J$ is less than $J_{\text{present}}$, the system will continue the remaining incomplete session and continues to compare the value of $J$ with $J_{\text{present}}$, until the two values are equal $J_{\text{present}}$. When the two values are equal, the measurement device plays a simple melody to indicate that the treatment course has been completed.

4. Experimental Results

In order to demonstrate the efficiency of the proposed system, some experiments were performed. First, we use the counterpoise to test the function of linearity of the pressure sensor modules. We allocate a counterpoise on the foot sole from 0kg to 2kg with a linear increment of 0.2kg. Figure 6 shows that the experimental data obtained from the pressure sensors is tallied with the true value. From the result, we can conclude that the sensor modules can perform well to measure the exact load. Furthermore, to investigate the practicality of the measurement system, two healthy heavy and light volunteers from the university participate in the experiment trial. These two volunteers weighted 92kg and 47kg, respectively. Figure 7 shows the weight measurement data obtained for volunteers walking with a gait cycle (1.2 sec for 92 kg and 0.8 sec for 47 kg) on a dynamical force plate and on wearing the proposed measurement outsole, respectively. Compared with dynamometer, the proposed system can perform in real trial case well within a reasonable error range.

To test the dynamical warning of alarm system, we record the real time plantar pressure of the volunteer with weight 92 kg continuously walking for 16 seconds. Two mechanisms of prophylaxes are given in this trial: (1) when the measured pressure exceeds the default pressure of 40 Kg, the alarm emits two short beeps informing the patient to increase the pressure; (2) when the measured pressure exceeds 80 Kg, the alarm emits a continuous beep to warn the patient that the loading is excessive. Figure 8 depicts the experimental results, there are 8 alarms indicating the over-loading and 65 alarms for under-loading.

5. Conclusions

This study has developed a microprocessor-based plantar pressure measurement and control
system. In this system, five pressure sensors are placed under the sole in regions corresponding to more important areas of pressure distribution. An acoustic alarm sounds when the foot loading exceeds the maximum permissible plantar pressure, thereby allowing a patient with a lower limb fracture to control the foot loading to a suitable level such that the efficiency of the callus recovery is optimized. The data obtained from the plantar pressure assessment instrument is stored into an EEPROM and provides a physiotherapist with valuable information to support his/her evaluations during the rehabilitation process. The experimental results have proven the accuracy of the proposed pressure measurement and dynamical alarm function of pressure control system.

6. References


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Table 1. Physical characteristics of pressure sensor.

<table>
<thead>
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<th>Name</th>
<th>specification</th>
<th>Name</th>
<th>specification</th>
</tr>
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<tbody>
<tr>
<td>Diameter</td>
<td>6mm</td>
<td>Safe excitation voltage</td>
<td>3V AC or DC</td>
</tr>
<tr>
<td>Thickness</td>
<td>0.6mm</td>
<td>Recommended excitation voltage</td>
<td>1V to 2V AC or DC</td>
</tr>
<tr>
<td>Weight</td>
<td>0.5g</td>
<td>Bridge resistance</td>
<td>120Ω±10%</td>
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<tr>
<td>Non-linearity</td>
<td>±1% RO</td>
<td>Safe temperature range</td>
<td>−20℃～70℃</td>
</tr>
<tr>
<td>Hysteresis</td>
<td>±1% RO</td>
<td>Rated capacity</td>
<td>500 kPa</td>
</tr>
<tr>
<td>Rated output</td>
<td>1mV/V ±20%</td>
<td>Classification</td>
<td>H</td>
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Fig. 1. Structure of proposed pressure measurement system.

Fig. 2. Photograph of hardware components of measurement system.
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Fig. 3. Sensor modules attached to outsole.

Fig. 4. Circuit of pressure sensor amplifier
Fig. 5. Software flowchart.
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Fig. 6. Linear properties of pressure sensors.

(a) The experiment for 92 kg.  
(b) The experiment for 47 kg.

Fig. 7. Force measurements during 1 gait cycle.
Fig. 8. Force measurements during extended walking.
Microprocessor-based plantar pressure measurement system for rehabilitation of lower limb.

以微處理機為基礎研製腳底壓力力量測系統——
應用於下肢復健

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摘要

對於下肢骨折病患於執行復健療程中，適當的下肢載重訓練有助於骨痂成形，進而促進下肢骨折的癒合，加速患者的復原。但是，復健的過程中，患者自身很難將載重控制在某一安全範圍內。因此，本文研究主旨為設計一套以微處理器為基礎具生物回饋之下肢載重回饋復健系統，讓患者可以藉此系統，在復健過程中的載重訓練中把載重控制在有效而安全的範圍內。本文提出的系統是以五個壓力感測器埋設於鞋墊上，量測腳底五個較重要的區域之壓力值，如腳後跟中心、蹠骨及腳拇指等…。本系統有兩項重要的特色：(1)在步態過程，當載重超過預設的限值時，控制器會發出聲響，藉此聲音的回饋作用，患者可以調整載重以控制在預期的載重範圍內。(2)記憶體裝置的設計，可將復健過程中的載重資料提供作為臨床分析。實驗結果得知，壓力感測器的線性度及人體實際量測均能符合預期之結果，確認此量測系統的可靠性。此結果可做為下一階段臨床上的應用。

關鍵詞：腳底壓力、壓力感測器、復健、微處理器

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