Design and Feasibility Study of Instrumented Shoes for Level Walking and Stair Ambulation

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Abstract
Analysis of foot-floor reactions during locomotion is helpful for clinical diagnoses of dynamic stability, walking ability, and balancing. This study integrates a pair of instrumented shoes and a microprocessor-controlled data logger to build a portable system for measuring vertical foot-floor reaction forces. A series of tests was performed to validate the accuracy of the instrumented shoe system in measuring ground-reaction forces (GRFs) of individuals during level walking, and stair ascent and descent. Plantar forces of 5 healthy young (HY) adults and a stroke patient (SP) with good mobility were measured continuously during level walking, stair ascent and descent for up to 10 min. Their temporal and spatial gait parameters were calculated. The SP exhibited substantially longer stride time, and double support and single support periods during stair ambulation than the HY group. Furthermore, the SP demonstrates bilaterally asymmetrical force waveforms, center of pressure (COP) loci and temporal gait parameters. Experimental results indicate that the proposed shoe system can acquire accurate GRF data for an extended period. Based on GRF data, the system can generate data for temporal gait parameters, and GRF waveform and cyclogram patterns for clinical diagnosis of pathological gaits.

Keywords—Stair, gait analysis, plantar force, instrumented shoes
1. Introduction

Walking is essential to daily living. Gait performance differs among physically challenged individuals. How to quantitatively and qualitatively assess and improve walking ability of people with disabilities is a significant problem in rehabilitation medicine. When evaluating patients, amounts of atypical loading or atypical loading patterns may reflect systemic or localized lower-extremity pathologies. Such analytical information may be indicative of or predictors for further pathology of an existing pathology [1]. Therefore, analyzing foot-ground reactions of persons with disabilities during daily locomotion can help doctors diagnose walking, stability, and balancing performance.

Information of foot-floor reactions during locomotion are widely applied in shoemaking, clinical diagnosis, sports medicine, rehabilitation engineering, orthopedics and biomechanics research. However, most gait studies were conducted in laboratories and relied on force platforms attached to fixed supporting surfaces to collected foot-floor reaction data. Such an experimental setup impedes the acquisition of successive gait data during one’s daily life. Numerous portable systems that measure plantar pressure have been developed to overcome this limitation (e.g., the Pedar insole system, F-Scan system, Parotec system, Musgrave footprint system, portable DynoGraphy system). The accuracy and reliability of these commercially available systems have been analyzed [2–6]. Although expensive, these plantar-pressure measurement systems offer clinicians a high degree of portability and permit use at several sites.

Utilizing portable equipment to measure gait performance has advantages, such as collecting onsite data during a subject’s daily activities over a prolonged period, which may provide many perspectives for clinical assessments. Several studies applied commercially available plantar-pressure systems to assess pathological gait during level walking [7–11], whereas other studies developed portable systems for measuring plantar forces of persons with disabilities during level walking [12–14].

In addition to level walking, many studies employed various motion-analysis systems combined with instrumented stairs to assess gait during stair ambulation. Most of these studies were conducted using normal subjects [15–22], and only a few investigated stair
climbing of patients with triple arthrodesis, subtalar fusion, knee implants, limb amputation or total hip arthroplasty [23-26]. Only one study utilized a portable measurement system to investigate the gaits of healthy adults during stair ambulation [27]. However, we are not aware of any quantitative study of the gait of people with a disability while ascending and descending stairs in real environment. A reliable portable and durable system is required to achieve this objective. This study presents a novel load-cell-based instrumented shoe system that measures plantar forces. The accuracy of the proposed system was verified in the laboratory via static and dynamic calibrations. System feasibility was evaluated by 5 healthy young (HY) adults and 1 stroke patient (SP) during successive level walking, and ascending and descending stairs.

2. Material and Method

2.1 Instrumented shoes design

An instrumented shoe consists of a forefoot part and rearfoot part. Each part has a surface and base plate made of 15mm-thick composite material. This material is lightweight and sufficiently firm that it can be stepped on. Four and 3 load cells were sandwiched between plates in the forefoot and rearfoot parts, respectively (Fig. 1). Each load cell is 9.5mm in height, 20mm in diameter, and 20g in weight. Each load cell is sunk in a 5mm-deep circular cavity on the surface plate. Additionally, each load cell withstands a 1.5mm-thick steel coin sunk in a circular cavity 5mm deep on the base plate. (Fig. 1). This design ensures that overall reaction force exerted onto surface plates is transmitted only to load cells. One load cell located at each rear shoe (heel position) can withstand 100kgw (LM-100KA, Kyowa, Inc., Japan); and the rest load cells each can withstand 50kgw (LM-50K, Kyowa, Inc., Japan). Seven load cells were placed in a shoe to cover the hallux, medial metatarsal, lateral metatarsal, medial midfoot, lateral midfoot, and calcaneus regions. In total, 14 load cells were used for each pair of instrumented shoes. Fishing line was used to stitch the surface and base plates together. A spring steel plate, adjustable in length, was used to join the forefoot and rearfoot shoe parts and allow flexion between these parts. The shoes resemble a pair of sandals with Velcro shoelaces for foot-width adjustment (Fig. 2a). One 7-channel signal pre-amplifying circuit board was built for each shoe using instrumentation amplifiers (INA122, Burr-Brown, Inc., USA). Amplifier gain was set to 553 by using 365Ω external resistors.
2.2 Data logger

A 720g-weight logger (170mm × 160mm × 5.5mm) had a battery and CompactFlash (CF) memory card that were used together with the instrumented shoes to record plantar forces (Fig. 2b). The data logger, powered by a 7.2V rechargeable Li-ion battery, had circuit boards for signal acquisition and digital control modules. An 8-bit microprocessor (C8051F020, Silicon Laboratories, USA) was employed to operate the logger.

The signal-acquisition module was powered at ±5V converted directly from the 7.2V Li-ion battery by a positive low dropout regulator (LT1117-5, Linear Technology Corp., USA). This module converted pre-amplified load cell signals of each shoe via a separate A/D converter. Each A/D converter (ADS8344N, Texas Instruments, Inc., USA) had an 8-channel multiplexer, 16-bit resolution, and a fixed sampling rate of 1000 sample/sec per channel.

The digital control module was powered at +3.3V provided by an LT1117-3 regulator, and at +5V provided by an LT1117-5 regulator with power converted from the 7.2V Li-ion battery. This module consists of digital I/O circuitries that control data storage, the LCD module, and RF synchronization. A CF memory card with a storage capacity of 512 Mbytes was utilized for data storage. A wireless transmitter set comprised of a 315MHz RF receiver module and decoder IC (PT2272, Princeton Technology, Taiwan) was used to register externally triggered events and synchronize the logger and video camcorder.

A 128×64 dot graphic LCD module (LMG-SSC12A64, SDEC Technology Corp., Taiwan) displayed the information from operational modes. With this information, users can control the logger, check signal quality, and manage data files stored on the CF card manually via control buttons.

2.3 System calibration
Both static and dynamic calibrations of the instrumented shoes were performed in the laboratory before the onsite experiment. Static calibration assured that no significant cross talk exists between load cells. A 20-pound static counterweight was applied directly above each load cell on the surface plate of each instrumented shoe. Dynamic calibration was performed to determine a set of offset and scaling parameters that converts corresponding voltage signals from load cells into engineering units of force (N). Dynamic calibration was conducted by an experimenter wearing the instrumented shoes and walking on a force plate system (BP2436, ATMI, Inc., USA). Data were collected synchronously by the shoe system and force plate system with an external trigger signal. According to the linear characteristic between loading force and output voltage of a load cell, overall plantar force can be represented in terms of load cell voltages and corresponding parameters:

\[
F_{\text{overall}} = \sum_{i=1}^{14} F_i = \sum_{i=1}^{14} s_i (V_i - V_{o,i})
\]  

(1)

where \(F_i\), \(s_i\), \(V_i\), and \(V_{o,i}\) are the load cell force, scaling parameter, output voltage signal and unloading voltage offset of the \(i\)th load cell, respectively. Unloading voltage offset of each load cell can be determined easily from data collected when no forces were exerted on the shoes. Since the overall plantar force and voltage signals of the 14 load cells were measured over time, the best set of scaling parameters was determined by minimizing the overall square errors:

\[
\min \sum_{i=1}^{n} \epsilon_i^2 = \min \sum_{i=1}^{n} \left( \tilde{F}_i - \sum_{i=1}^{14} s_i (V_{i,t} - V_{o,i}) \right)^2
\]

(2)

where \(\epsilon_i\) is error and \(\tilde{F}_i\) is the resultant force measured by the force plate system at time \(t\).

Dynamic calibration was conducted repeatedly over 2 weeks to achieve consistent system performance. The derived scaling parameters were saved for subsequent applications.

2.4 Onsite gait measurement

2.4.1 Subjects
Five healthy male adults, aged 25–27 years (weight, 73.0±17.2 kg; height, 174.6±7.5 cm) and 1 male SP, aged 24 years (weight, 60 kg; height, 167 cm), were recruited for the feasibility test. The SP was recruited from the rehabilitation department at Chang-Gung Memorial Hospital. A medical doctor was responsible for screening the stroke subject and confirmed that the subject was capable of ascending and descending stairs without assistance in daily life. The SP had right temporal arteriovenous malformation (AVM) with intra-cerebral hemorrhage (ICH). He underwent a craniotomy with excision of the AVM and ICH at onset, and participated in this study 2 years after stroke onset. Clinical manifestations were left hemiplegia and sensory impairment. The patient underwent motor recovery assessment at the time of gait analysis. Motor recovery was assessed using the six stages of Brunnström’s recovery [28]. Ankle and foot movement were also graded into 6 stages using similar principles. The 6 grades for ankle and foot movement were as follows: (1) flaccidity; (2) synergy development (minimum ankle or toe movement); (3) mass ankle inversion without voluntary toe extension; (4) ankle eversion and semivoluntary toe extension with a small range of motion (ROM); (5) mass toe flexion and extension (variable ROM); and, (6) individual toe movement (full ROM). Brunnström’s recovery stages 3 and 4 indicate high synergistic and mass movements, whereas Brunnström’s recovery stages 5 and 6 indicate isolated and selective movement. The lower-limb motor recovery of this subject was stage IV on proximal parts and stage II on distal parts in Brunnström’s recovery staging. This study was approved by the Institution Human Ethics Committee. After an explanation of this study, all subjects provided informed verbal consent.

2.4.2 Experimental protocol

All subjects wore the instrumented shoes and carried the logger in a small knapsack. Subjects were instructed to walk on level surface for 25 m and then climb stairs 18.5 cm high and 27.5 cm deep at their own pace. According to his ability and comfort, the stroke subject could choose to stop climbing after he had climbed >11 stairs (half of a floor). The control group had to climb at least 77 stairs (3.5 floors). After a short break, subjects were instructed to walk down the stairs and return to the task start position. Each task took approximately 10–20 minutes from the time the subject started to walk on the level surface. Two experimenters, one nurse and one physical therapist, walked aside and behind the SP to prevent the subject from falling. The hemiplegia subject was allowed to place his sound hand on a handrail or wall for
support. This subject first used his sound foot to step on an upper stair when ascending and his affected foot on the lower stair when descending. During the task, an experimenter videotaped subject movement and used a laser pointer with a wireless transmitter to synchronize the video and logger.

2.4.3 Data analysis
Logait analytical software was programmed by LabVIEW 7.0 (National Instruments Ltd., USA) and used to calibrate signals, synchronize video, and segment and process data (Fig 3). This software is equipped with an arithmetic core of Calculate Express 2.4 for calculating maximum peak ground reaction force (GRF), temporal parameters, and cyclogram patterns of each gait cycle (GC).

The temporal stride parameters, GRF waveform and cyclogram patterns were analyzed. The GRF data for a subject obtained by the shoe system was segmented into datasets for level walking, ascending stairs, and descending stairs. In each condition, a GC was defined as the period from a foot landing to its next landing, and was further divided into single support (SS) and double support (DS) periods. Both support periods were normalized to stride time and presented as percentage of GC (%GC). In each GC, peak GRF of each foot was calculated and normalized to subject body weight (%BW). A cyclogram for each walking condition was derived from individual weight bearing and supposititious position of the 14 load cells by calculating overall center of pressure (COP) [9]. The analysis of COP data was conducted based on subjective observations. The level walking speed of a subject was estimated by the time he finished walking 25m. Bilateral differences in the SS period and peak GRF were tested using a t-test. To reduce the inconsistency in gait when during turning around, speeding up or slowing down, only 5 middle gait cycles per half floor (11 stairs) were selected for data analysis during stair ambulation. In total, 20 GCs during steady level walking and 20 GCs during steady stair ambulation were selected for data analysis to reduce inter-subject variability possibly caused by fatigue.

Figure 3. Logait software shows the synchronized load cell data and video image of the stroke patient
3. Results

3.1 System accuracy

No significant cross talk of load cell signals existed in static calibration. The architecture of the instrumented shoe ensures that each load cell shares the weight applied to the plate surface. The calibration result also ensures that the arithmetic sum of weight on load cells in a shoe closely represents overall weight bearing of the foot. The averaged root mean square of static calibration errors was 0.1%. Figure 4(a) shows a typical result of load cell signals measured during a static calibration task in which the arithmetic sum of weight on load cells was compared with the force plate measurement.

Over a 2-week test period, dynamic calibration derived consistent gains for each load cell with coefficients of variance of 0.02–0.07. The shoe system had a 1.72% averaged root mean square dynamic error compared to that of the force plate system. Figure 4(b) shows a typical result of the dynamic calibration task in which the arithmetic sum of weight on load cells was compared with the force plate measurement.

Figure 4. Typical (A) static and (B) dynamic calibration results of overall GRFs measured by force platforms and derived from load cell data.

3.2 Level walking and stair ambulation

Table 1 presents the gait parameters of 5 HY subjects and the SP. The SP walked on level surface at a slower speed than the HY subjects (SP=60.8 cm/s; HY=96.1±9.4 cm/s). The gait parameters in level walking of the SP were similar to those of the HY subjects (Table 1). Experimental results demonstrate that the SP had good mobility during level walking and could ascend 77 stairs. The SP took one step per stair when ascending and descending stairs with his sound hand on the rail (ascending) or wall (descending) for support. All HY subjects also took one step per stair during stair ambulation without any support. During stair ambulation, the SP had longer stride time and time spent on each stair than the HY subjects, and had a reduced SS period in his affected leg, approximately 10%GC, compared with that of the HY subjects (Table 1). The average stride time of the SP while ascending and
descending stairs compared with level walking was increased by 1.6 and 1.8 sec (123% and 140%), respectively; however, average stride time did not change significantly for HY subjects (Table 1).

All healthy subjects demonstrated ‘M’ shaped double humps and bilaterally symmetric GRF waveforms, and symmetric butterfly-shaped cyclograms during level walking and stair ambulation (Figs. 5 and 6). The SP applied 10–30%BW reduced peak GRF than HY subjects in his affected leg when ascending and descending stairs. In short, side differences (sound vs. affected leg) in the SP were enlarged when level walking changed to stair ambulation. The SP had a peak sound GRF of 121.2%BW and 117.7%BW while ascending and descending stairs, respectively, those values were approximately 9 %BW increased from that for level walking (mean, 108.4%BW). Comparatively, the peak GRF of HY while ascending and descending stairs increased approximately by 12 and 33%BW, respectively, from that for level walking (mean, 107.3%BW) ($p<0.007$) (Table 1). Peak force was significantly greater than that for level walking for HY subjects during the push-off phase when ascending stairs (2nd hump peak, average 118.4 %BW) and when the foot landed while descending stairs (1st hump peak, average 139.9% BW) (Table 1). Unlike the HY subjects, the SP had a similar magnitude of peak GRF (range, 103.3–109.1%BW) in his affected leg for all walking conditions.

The HY subjects had shorter COP excursions during stair ambulation than during level walking (Fig. 6). However, GRF waveforms of the SP varied and were bilaterally asymmetrical for all walking conditions (Fig. 5). His cyclogram patterns had triangular shapes with smaller COP excursion in the affected foot than that for the non-affected leg (Fig. 6) The SP employed longer COP excursions in his sound limb while keeping COP of the affected limb in a pivot spot during ambulation. According to the cyclogram loci, the main weight-bearing area of the foot was shifted to the forefoot while descending stairs and was centralized at the midfoot in HY subjects while ascending stairs. Nevertheless, the cyclogram patterns of the SP while ascending and descending stairs varied and deviated from that of the HY subjects.
4. Discussion

With a sturdy and durable structure design, the shoe system had adequate accuracy and reliability according to calibration results. The shoe system was tested and proved capable of measuring temporal gait characteristics and acquiring GRF data during daily life. In another study, the proposed system was successfully employed to assess level walking and stair ambulation of several SPs with different levels of disability. These experimental results are to be documented elsewhere [29]. This study demonstrates the capability of the proposed shoe system to continuously acquire plantar forces, and analyze each gait cycle during ambulation. System cost is estimated at US $7000, which is cheaper than most commercially available plantar pressure systems.

Experimental results suggest that the proposed shoe system acquired data accurately. This was demonstrated by the fact that all HY subjects exhibited bilaterally symmetrical gait patterns and moderately slow walking speeds. The average level walking speed of the HY subjects was 96.1 cm/s, about 10–30% slower than those for free walking in previous studies (range, 106–139 cm/s) [16, 30]. Therefore, the derived cycle times in all test conditions were 10–20% longer than those reported by Nadeau et al. and Riener et al. [16, 18]. Similarly, the SS (swing) period of the HY subjects in this study was approximately 80% of those reported by Nadeau et al. and Riener et al. [16, 18]. It is possible the subjects were unfamiliar with the instrumented shoes and testing environment, and adopted a shorter step length resulting in slower walking speeds. The unfamiliarity of the shoe system and test environment may have caused subjects to adopt different gait patterns. Further investigation is required to verify possible effects caused by the proposed shoe system on gait. Despite this restriction, this study convincingly demonstrated several of the temporal stride characteristics and gait patterns during walking and stair ambulation. Christina et al. reported that HY adults have an
average 140–148%BW for the 1st peak GRF while descending stairs [15]; this range was similar to that identified in this study (Table 1). The average level walking speed of the SP was 60.8 cm/s. People who suffered hemiplegic strokes have fast, medium and slow walking speeds of 63, 41, and 25 cm/s, respectively [31].

Experimental results also indicate that the proposed shoe system has good ability to distinguish between pathological gaits. The SP was age-matched with the HY subjects, and had good stamina and consistency during walking and stair ambulation. Despite the relatively similar cycle time in level walking for the HY group, the SP had a longer DS period and shorter SS period in his affected leg than those of HY subjects during stair ambulation. Additionally, experimental data show that the SP applied less peak GRF in the affected leg while ascending stairs and less peak GRF in both legs while descending stairs than HY subjects. Unlike the HY subjects, all GRF waveforms of the SP were bilaterally asymmetrical under all walking conditions. Previous studies also demonstrated that GRF waveforms of stroke subjects varied in shape during level walking [32–34]. In addition to distinguishable difference in gait parameters between the SP and HY subjects, asymmetrical cyclogram patterns of the SP had a shorter COP excursi on in the affected foot during all walking condition. Previous studies also showed that cyclogram patterns of stroke patients differ during level walking [9]. The SP had longer COP excursion in his sound limb while keeping the COP of the affected limb in a pivot position during level and stair ambulation. The reasons may be that stroke subjects rely on the sound limb for breaking, support and propulsion, thus preserving the function of the heel strike and push off phases during level and stair ambulation.

In this preliminary study, experimental results demonstrated that differences in gaits between the PS and HY subjects can be distinguished according to gait parameters, even though the SP in this study was young and capable of stair ambulation during daily living. Moreover, with this proposed shoe system, walking ability and prognostic progress of stroke victims can be documented via various outcomes such as temporal gait parameters, bilateral symmetry of plantar force waveforms, peak weight-bearing, and COP loci of cyclogram patterns. Symmetry and consistency are also important mobility factors in gait analysis, both of which require an assessment tool, such as the proposed shoe system, that is reliable and has the
capacity for prolonged data collection. Kim and Eng determined that symmetry GRF during level walking was correlated with symmetry in temporal, but not distance, gait measures among 28 stroke patients [35]. However, no study analyzed the gait of people with a disability while ascending and descending stairs in real environment. The proposed shoe system overcomes environmental limitations when measuring gait performance in ambulation tasks other than level walking for an extended period. Furthermore, specific physical conditions, such as electrocardiography (EKG) and electromyography (EMG) can be monitored using a conventional portable instrument. For instance, a multi-transducer logger can be synchronized with this proposed shoe system to record muscle activity, joint posture, or movement speed of the lower extremities [36].

This study was restricted by the small numbers of subjects and employed only HY subjects to validate the shoe system accuracy and reliability. Single shoe design cannot possibly fit foot sizes and shapes of every variety. Moreover, factors including unfamiliarity of the shoe system and test environment, foot-width and shape of individual subject, and handrail/wall use by the SP may affect the outcome of measurement. These factors were not addressed in this study, making the exact effect of a particular factor on the experimental observations unclear. Further experimental design is needed to uncover how individual factors might affect the result of gait measurement.

Despite small number of subjects in this study, experimental results obtained by this feasibility study indicate that the proposed shoe system is useful when assessing pathological gait and is an alternative solution for evaluating gait performance of physically challenged individuals.

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References


**Figure captions**

Figure 1. Structural view and sensor locations of an instrumented shoe

Figure 2. Outward appearance of the (a) instrumented shoes and (b) data logger

Figure 3. Logait software shows the synchronized load cell data and video image of the stroke patient

Figure 4. Typical (A) static and (B) dynamic calibration results of overall GRFs measured by force platforms and derived from load cell data

Figure 5. Typical GRF waveforms of a HY subject (left) and the SP (right) subject during (a) level walking (b) ascending stairs and (d) descending stairs (dark lines indicate the right or the sound foot; grey lines indicate the left or affected foot)

Figure 6. Typical cyclogram patterns of a HY subject (top row) and the SP (bottom row) subject

**Table captions**

Table 1. Gait parameters of the SP and HY subjects during level and stair ambulation
Table 1. Gait parameters of the SP and HY subjects during level and stair ambulation, and related results from reference studies

<table>
<thead>
<tr>
<th>Gait parameter</th>
<th>HY (n=5)</th>
<th>SP</th>
<th>HA[^16]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total stairs ascended</td>
<td>88</td>
<td>77</td>
<td>n.a.</td>
</tr>
<tr>
<td>Walking speed (cm/s)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Level</td>
<td>96.1±9.4</td>
<td>60.8</td>
<td>116[^16]</td>
</tr>
<tr>
<td>Manner (steps/stair)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ascent</td>
<td>1</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>Descent</td>
<td>1</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td>Stride time (sec)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Level</td>
<td>1.42±0.11</td>
<td>1.32±0.09</td>
<td>1.15[^16], 1.11[^18]</td>
</tr>
<tr>
<td>Ascent</td>
<td>1.53±0.17</td>
<td>2.95±0.25</td>
<td>1.41[^18]</td>
</tr>
<tr>
<td>Descent</td>
<td>1.44±0.06</td>
<td>3.18±0.32</td>
<td>1.30[^16], 1.19[^18]</td>
</tr>
<tr>
<td>Stair duration (sec/stair)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ascent</td>
<td>0.76±0.09</td>
<td>2.95±0.25</td>
<td></td>
</tr>
<tr>
<td>Descent</td>
<td>0.72±0.03</td>
<td>3.18±0.32</td>
<td></td>
</tr>
<tr>
<td>Double support (%GC)</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Level</td>
<td>39.7±2.9</td>
<td>42.8±3.4</td>
<td>26.0[^16], 27.2[^18]</td>
</tr>
<tr>
<td>Ascent</td>
<td>43.8±5.4</td>
<td>56.0±2.5</td>
<td>27.2[^18]</td>
</tr>
<tr>
<td>Descent</td>
<td>39.3±2.6</td>
<td>50.8±4.0</td>
<td>20.6[^16], 22.4[^18]</td>
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<tr>
<td>Single support (%GC)</td>
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<td></td>
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<tr>
<td>Level</td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Sound (R)</td>
<td>29.1±0.8</td>
<td>31.5±4.4*</td>
<td>74[^16], 77.8[^18]</td>
</tr>
<tr>
<td>Affected (L)</td>
<td>30.0±2.9</td>
<td>25.8±3.2</td>
<td></td>
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<tr>
<td>Ascent</td>
<td></td>
<td></td>
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<tr>
<td>Sound (R)</td>
<td>28.7±3.0</td>
<td>25.9±2.4*</td>
<td>72.8[^18]</td>
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<tr>
<td>Affected (L)</td>
<td>28.9±2.8</td>
<td>18.1±1.2</td>
<td></td>
</tr>
<tr>
<td>Descent</td>
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<tr>
<td>Sound (R)</td>
<td>31.6±2.1</td>
<td>29.9±3.6*</td>
<td>79.4[^16], 77.6[^18]</td>
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<tr>
<td>Affected (L)</td>
<td>29.8±1.3</td>
<td>19.3±1.7</td>
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<td>Peak GRF (%BW)</td>
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<td>Level</td>
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<tr>
<td>Sound (R)</td>
<td>107.3±1.7</td>
<td>108.4±5.3*</td>
<td></td>
</tr>
<tr>
<td>Affected (L)</td>
<td>106.8±1.4</td>
<td>103.3±3.5</td>
<td></td>
</tr>
<tr>
<td>Ascent</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sound (R)</td>
<td>118.6±1.9</td>
<td>121.2±11.9*</td>
<td></td>
</tr>
<tr>
<td>Affected (L)</td>
<td>118.2±0.7</td>
<td>108.3±7.7</td>
<td></td>
</tr>
<tr>
<td>Descent</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sound (R)</td>
<td>140.8±15.0</td>
<td>117.0±6.6*</td>
<td>140-148[^15]</td>
</tr>
<tr>
<td>Affected (L)</td>
<td>139.0±13.4</td>
<td>109.1±6.1</td>
<td></td>
</tr>
</tbody>
</table>

Mean ± s.d. values of the SP subject were computed by 20 steps.

* Significant difference between sound side and affected side (p<0.001, t-test)
Figure 1. Structural view and sensor locations of an instrumented shoe

Figure 2. Outward appearance of the (a) instrumented shoes and (b) data logger
Figure 3. Logait software shows the synchronized load cell data and video image of the stroke patient.

Figure 4. Typical (A) static and (B) dynamic calibration results of overall GRFs measured by force platforms and derived from load cell data.
Figure 5. Typical GRF waveforms of a HY subject (left) and the SP (right) subject during (a) level walking (b) ascending stairs and (d) descending stairs (dark lines indicate the right or the sound foot; grey lines indicate the left or affected foot)
Figure 6. Typical cyclogram patterns of a HY subject (top row) and the SP (bottom row) subject.