LOWER EXTREMITY AMBULATORY FEEDBACK SYSTEM FOR PEOPLE WITH AMPUTATIONS GAIT ASYMMETRY TRAINING

by

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ABSTRACT

A wireless, wearable, real-time gait asymmetry detection system—the Lower Extremity Ambulatory Feedback System (LEAFS)—has been validated by comparison to clinical motion capture (force plate and three-dimensional cameras) measurements, and evaluated in training sessions with seven subjects. LEAFS is a low-cost in-shoe gait detection device that provides real-time auditory feedback based on stance time ratio and allows long-term gait asymmetry training to be performed outside of the clinical environment. Stance time ratio, which is also known as Symmetry Ratio (SR), is calculated by dividing the stance time on one limb (typically the more affected limb) by the other, and control subjects have been shown to have SR of 1.02 ± 0.02. The validation test results demonstrate that the SR measured by LEAFS as compared to clinical motion capture results has a mean error of 0.003 ± 0.05 for control subjects and 0.008 ± 0.04 for subjects with unilateral trans-tibial amputations. The LEAFS was used for gait asymmetry training in seven subjects with unilateral trans-tibial amputations; subjects received six 30-minute training sessions over a 3-week training period. The results demonstrate that LEAFS is accurate at measuring mean SR of a trial of steps, and it is reliable and practical to use LEAFS to train the gait of patients with unilateral trans-tibial amputations by bringing their SR towards a normal range.

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Lastly, I offer my regards and blessings to all the patients and participants in this project, and all those who supported me in any respect during the completion of this thesis.

CHAPTER 1

INTRODUCTION

Gait asymmetry occurs commonly among subjects with Parkinson's disease (PD), or among people with lower-limb amputation, etc. [1][2]. This can lead to inefficient gait, poor balance, and higher metabolic cost, thus resulting in joint over-use, osteoarthritis, and higher possibility of falling, which lead to lower quality of life, or even death [3][4][5][6][7][8][9] [10][11][12][13][14]. Further, gait asymmetry causes an abnormal gait pattern, which indicates disability, and might cause negative psychological impact on the patient.

Among patients with lower-limb amputations, these effects become more obvious. Patients with amputations tend to use their intact leg more than the prosthetic leg due to discomfort, and thus the intact leg is more likely to develop osteoarthritis and joint damage [15]. Meanwhile, the patients' gait pattern changes and causes other negative effects [11][12][13][14]. Our hypothesis is that, because of the lack of proprioceptive feedback from the prosthetic leg, the patient cannot feel the force distribution between the prosthetic leg and the intact leg, and thus cannot adjust force and posture as well. This leads to a different balance pattern and asymmetric gait. Thus it is possible to reduce or eliminate gait asymmetry by offering real-time feedback to the patient during walking.

1.1 Existing devices and research

Some existing devices and research products are introduced in the following sections, including laboratory based systems and some commercialized patient mounted systems.

1.1.1 Laboratory based systems

Here we will introduce some existing laboratory based systems and explain the pros and cons of those systems.

1.1.1.1 Force mat

One widely used gait analysis tool is force mat, such as GAITRite from CIR Systems, Inc. [16]. A force mat is a mat embedded with force sensors that can measure GRF while the subject is walking on the mat. GAITRite has more than 18,000 sensors arranged in a 48×288 grid and can collect data at 60Hz, 120Hz, or 240Hz. The length of mat ranges from 12 to 26 feet. This type of system requires special training for data analysis, and can be used to gather force data from only a limited number of steps due to the size limit of the mat, and thus is not suitable for continuous data collection or outdoor use.

1.1.1.2 3D motion capture with force plate

Clinical motion analysis laboratories typically contain a force plate(s) combined with a video motion capture system using reflective markers. Kistler force plate [17] and AMTI force plate [18] are some of the most frequently used force plates. Motion capture/analysis systems include those from companies such as Vicon [19], and Motion Analysis Corp. [20]. These systems are highly accurate but also highly expensive and are usually found in hospital gait labs, or motion studios in the movie industries. They are extremely expensive and not suitable for home use.

1.1.1.3 Treadmills with force plates

The previous systems gather information from a limited number of steps, or from limited ground contact pattern information, such as from a force mat. However, the analysis of gait asymmetry requires a comparison of the walking pattern of both limbs over a large sample size yielding more complete ground contact force information that will make comparisons more accurate and reliable. Treadmill and force plate based systems are designed to allow subjects to walk continuously with their gait continuously measured. This allows the researchers to gather information from a large number of steps. One commercial product is CCF Treadmill [21].

A treadmill-based system with a force plate, developed by Edward and others [4], can measure several gait parameters such as stance phase, swing phase, and GRF at 50 Hz or higher. Another treadmill-based system, developed by Junho Park and others [22], used a laser sensor array and magnetic sensors to detect the foot position while subjects are walking on a treadmill. However, treadmill-based systems cannot be used to evaluate gait in the outdoor environment, and the subjects' movements were limited to a treadmill waking pattern as well.

1.1.2 Patient mounted systems

Here we will introduce some existing commercialized systems and explain the pros and cons of those systems.

1.1.2.1 Foot switches

Foot switches are one of the earliest existing devices for gait measurement. The Portable Gait Analysis Stride Analyzer from B & L Engineering, Tustin, CA, USA [23], uses insoles with pressure switches embedded in them. When using the system during a gait cycle, different switches are triggered at different times; thus the foot-floor contact pattern will be recorded for gait analysis. Foot switches are usually low-cost with a price from less than a hundred to a few hundred US dollars (the software and analysis equipment are included, the cost might reach to around 10k USD, depending on the manufacturer). The signal is easy to process since the readings are 1 or 0 for a switch. But these systems are usually not suitable to measure accurate GRF patterns, which requires force sensors that provide more than just an on and off measurement. The foot-floor contact pattern measured by foot switches may also vary over different subjects, since the position of switch under the foot may change according to the size and shape of foot.

1.1.2.2 Insole GRF detection systems

Since one of the most import parameters of gait that an in-shoe sensor system can measure is the GRF, there are similar insole systems embedded with force sensors, instead of switches. Different kinds of force sensors can be used, and based on the number, accuracy of sensors embedded in one insole, the system can cost from a few hundred dollars to over 10k USD. Some commercially available insole sensor devices that are used in clinical study or treatment are PEDAR from Novel Electronics, Inc [24]; F-Scan from Tekscan, Inc [25]; Parotec in-shoe pressure measurement system from Paromed Vertriebs GmbH & Co. [26]; CDG Ultraflex Gait analysis system from INFOTRONIC Medical Engineering [27]. Among these, PEDAR and F-Scan are the most frequently used devices, and have been evaluated in several publications [28][29][30][31]. The PEDAR system uses a force sensor matrix that contains 256 to 1024 sensors in one matrix [24], and has been shown to be highly accurate in detecting stance time and GRF compared to Kistler force plate over long time usage, with a sampling rate of around 100 Hz [31]. The system can collect and store data wirelessly for around 5 hours, and can be used in an outdoor environment [31]. The F-Scan system uses force sensors that can be trimmed to fit different size of shoes, with 960 to 1848 sensor cells per side, and the sampling rate can be up to 750 Hz, but a cable that is connected to a data storage device that was mounted on the user's waist, or a computer is required when using the system [25].

CDG sensor shoes with Bio Feedback Unit (BFU) can be used to detect underload and overload state under a certain area of foot, for each step separately during walking [27]. If the force under certain area exceeds a certain pre-set level, an acoustic signal sounds, and the location of over/under load is shown on an LED display. This sensor shoe is worn outside of the patient's shoe. Another kind of CDG sport shoes has sensors integrated into the shoe.

These systems have a high accuracy and can measure multiple gait parameters such as force distribution, stance/swing time, peak forces, etc., and can be combined with other motion capture tools such as video. But they are often prohibitively expensive for individual patients or small clinics to purchase; with a cost over 10k USD according to different system models and software packages. These systems are designed for full analysis of foot contact force, instead of evaluating gait stance time asymmetry to provide feedback for patients with gait asymmetry.

1.1.2.3 Other systems under research

Research involving gait pattern detection and feedback has a long history, as early as 1974 [32], when Donald Endicott and others developed a pressure sensor used for a leg load warning system.

A portable wearable sensor measurement system for walking [33] used 2 accelerometers to record data on a memory card at 50 Hz. This system can measure acceleration of the hip while walking. Another device [34] also used accelerometers to measure hip acceleration, and used acceleration ratio as verbal feedback to correct subjects' gait.

Current research involving insole force sensor systems for gait evaluation includes a wireless system for gait and posture analysis [35] using a 24-sensor insole pressure system based on hydrocells by Paromed [26]. This system can measure GRF from both feet at 80 Hz continuously for 5 hours, but is not specifically designed to evaluate gait asymmetry and does not give feedback to patients. Gait measurement devices that used multilayer insoles as force sensors [36][37][38] have a problem of drift on sensor readings, and other devices have problems such as unsuitable for long-term use due to large device size or low sampling rate [39][40]. Previous research done by Dr. Stacy Bamberg on gait shoe motion analysis instruments used multiple sensors but was not specifically designed for measuring gait asymmetry [41][42][43], and lacked feedback features.

1.2 LEAFS

The LEAFS system is designed to help patients using the feedback system on the computer to compensate for the lack of proprioceptive feedback of the force distribution under the prosthetic foot. The LEAFS system monitors the patients' gait asymmetry by acquiring the stance time difference between the prosthetic and intact limb. Once the stance time ratio, know as symmetry ratio (SR), passes the pre-set value, a beeping warning sound will be given to the patient. This system can be connected wirelessly to a computer, thus can be used both indoors and outdoors.

1.3 Contributions

The specific contributions of the author to the LEAFS project are:

1. Development of a fully working second version of the LEAFS;
2. Contributions to the third version of the LEAFS, particularly in sensor layout;
3. Validation of the LEAFS project symmetry ratio (including data collection and analysis);
4. Evaluation of the LEAFS ability to improve symmetry ratio in subjects with amputations (including data collection and analysis).

These are discussed in further detail in the context of this thesis.

1.3.1 First and second version of LEAFS

LEAFS was first developed by Dante Bertelli, a previous student in the BioInstrumentation lab of University of Utah. Based on the first version of this system, several changes and improvements have been made, and the second version of LEAFS has been developed and manufactured in this thesis project. A verification experiment as well as clinical effectiveness experiments of LEAFS have been done in this project as well.

The first version of LEAFS has two force sensing resistor (FSR) in each insole, and the data rate of the system was around 10-30 Hz. The system was using 7V batteries, and can continuously send and collect data for about 2 hours on two new batteries. In the second version of this system, the data rate was increased to around 200 Hz, and each insole has seven FSR, where six were used for calculation, The system used 3.7V batteries and can continuously send and collect data for about 12 hours on two new batteries. Also, new LabView based application software has been created as well [44][45][46][47]. The second version of LEAFS is discussed in detail in Chapter 3.

1.3.2 Third version of LEAFS

A third version of LEAFS was developed in this research, as a combination work with another Ph.D. research project, Phil Dyer's research on balance and stability in the BioInstrumentation lab. The third version of LEAFS used a different circuitry and software system, and overall reliability and accuracy of the system was improved. A total of 10 FSRs were used in this version of system, where nine were used for calculation [48][49][50]. Since the sensor layout is critically important to the accuracy of measurement, this specific topic is discussed in Chapter 3.

1.3.3 Clinical experiments' data collection

Two types of experiments were conducted in this research project: the system evaluation test as compared with a force plate and the clinical test of LEAFS. The system evaluation test was conducted in the MOCAP lab of University of Utah [51], the measurement of stance time and stance time ratio were gathered and evaluated from both LEAFS and force plates. The clinical effectiveness experiment was conducted in the University Hospital Rehabilitation Clinic; seven subjects with amputations were recruited in this experiment. The author of this thesis was in charge of the data collection of these experiments.

1.3.4 Data analysis

LEAFS data processing is solely completed by the author, using MATLAB, this involved the comparison analysis between LEAFS data and MOCAP force plate data, and analysis of the data from the clinical effectiveness experiment. The results are discussed in detail in Chapter 3.

CHAPTER 2

ABSTRACT PRESENTED AT AMA-IEEE 2010

The abstract [52] presented at the First AMA-IEEE Medical Technology Conference on Individualized Healthcare, Washington DC, March 22-23, 2010. Reprinted with permission from Linfang Yang and Stacy J. Morris Bamberg.

A Wearable Wireless Auditory Feedback System for Gait Rehabilitation

Linfang Yang, Philip Dyer, and Stacy J. Morris Bamberg

Introduction

Every year, 156,000 individuals in the USA lose a limb, and about half have a lower-limb amputation. For those who use a prosthetic limb, they will face a challenge in their future everyday lives: adapt to a new walking pattern. The common incidence of a new asymmetric walking pattern [1] can have serious health consequences, because in addition to altering the appearance of walking, asymmetry also results in increased metabolic costs and a higher incidence of knee osteoarthritis [2, 3]. Stance time ratio (the stance time of the affected limb divided by that of the normal limb) is recommended for evaluating asymmetric gait, and healthy adults have a typical stance time ratio of 1.02±0.02 [4].

Materials and Methods

Our instrumented insole system has been developed to wirelessly collect information pertaining to the forces underneath the feet during gait. A custom MATLAB® program analyzes the data in real-time, to create a Lower Extremity Auditory Feedback System (LEAFS) to provide an at-home option for gait rehabilitation (Fig. 1). Inputs include identifying the affected limb, selecting an asymmetry threshold, and type of auditory feedback.

Fig. 1. LEAFS: a) Interface, b) on intact and prosthetic limbs

Initial contact and final contact for each limb are identified in real time, to calculate stance times. The ratio of stance times is calculated following final contact on the affected limb. If the ratio is below the set asymmetry threshold, a loud beep occurs immediately; if the ratio is above the threshold, no beep occurs. *Our hypothesis is that using the LEAFS for training will allow a user to gradually reduce or eliminate gait asymmetry.*

We are conducting a prospective analysis of 10 subjects with gait asymmetry secondary to unilateral trans-tibial amputation. Individuals are evaluated in a motion analysis lab before and after six 30-minute training sessions with the LEAFS; post-tests will occur one week and six weeks after the final training session to investigate the persistence of improvements in gait symmetry. A physical therapist oversees the training and sets the threshold for feedback.

Results

Our first subject is currently undergoing training with the LEAFS. Fig. 2 shows three consecutive one-minute excerpts (*A*, *B*, and *C*) taken from the middle of the subject's third training session.

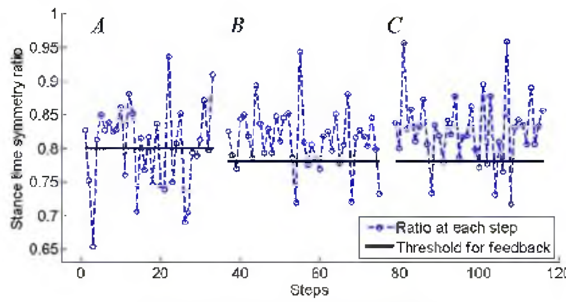

Fig. 2. One-minute excerpts during training

The importance of an appropriate asymmetry threshold is apparent. In *A*, the threshold is too high, and the minor adjustments made by the subject do not result in improvement. In *B* and *C*, the threshold is within an attainable range, and the mean symmetry increases as the subject works to alter her gait (Table 1).

Table 1. Changes in Symmetry During Three Training Excerpts

Excerpt	Mean Symmetry Ratio	Std. Dev of Symmetry Ratio
A	0.80	0.06
B	0.81	0.04
C	0.83	0.05

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CHAPTER 3

LOWER EXTREMITY AMBULATORY FEEDBACK SYSTEM

The paper on the following pages is in preparation for submission to the Journal of Rehabilitation Research and Development. The article title is Gait Asymmetry Training for Persons with Trans-Tibial Amputations using a Lower Extremity Ambulatory Feedback System. Contributing authors are Linfang Yang, Philip Dyer and Stacy Bamberg.

3.1 Abstract

A wireless, wearable, real-time gait asymmetry detection system—the Lower Extremity Ambulatory Feedback System (LEAFS)—has been validated by comparison to clinical motion capture (force plate and 3D cameras) measurements, and evaluated in training sessions with seven subjects. LEAFS is a low-cost, in-shoe gait detection device that provides real-time auditory feedback based on stance time ratio and allows long-term gait asymmetry training to be performed outside of the clinical environment. Symmetry Ratio (SR) is calculated by dividing the stance time on one limb (typically the more affected limb) by the other, and control subjects have been shown to have SR of 1.02 ± 0.02. When compared to clinical motion capture results, the validation test results measured by LEAFS have a mean SR error of 0.003 ± 0.05 for control subjects and 0.008 ± 0.04 for subjects with unilateral trans-tibial amputations. The LEAFS was used for gait asymmetry training in seven subjects with unilateral trans-tibial amputations; subjects received six 30-minute training sessions over a 3-week training period. The results demonstrate that LEAFS is accurate at measuring mean SR of a trial of steps, and it is reliable and practical to use LEAFS to train the gait of patients with unilateral trans-tibial amputations by bringing their SR towards a normal range.

3.2 Introduction

The quality of gait is closely related to quality of life, and one of the important features of gait is symmetry, since gait asymmetry is often considered an indication of gait pathology [1]. Gait asymmetry occurs commonly among people with Parkinson's disease (PD), stroke, cerebral palsy, hemiplegia, ligament deficiency, and lower-limb amputation, as well as other populations [1][2]. The consequences of gait asymmetry are harmful and can lead to poor balance, higher metabolic costs, joint over-use, osteoarthritis, and higher possibilities of falling, which can result in serious injury, or even death [40][4][5][6][11][7][8][9]. Further, gait asymmetry causes an abnormal gait pattern, which is a visible indication of disability, and can have a negative psychological impact on the patient. This research focused on developing a Lower Extremity Ambulatory Feedback System (LEAFS) to help reduce gait asymmetry.

3.2.1 Gait asymmetry studies background

There are over 2.5 million amputees in the United States, with 185,000 individuals in the USA undergoing amputation yearly [8]. Studies show that there is an increased chance of joint pain and degeneration as well as higher risk of osteoarthritis among population with amputations, complications that have been related to greater Ground Reaction Forces (GRF) and gait asymmetry [11][7]. Gait pattern changes may have negative effects on joints and the back [10]; up to 71% of unilateral lower limb amputees have reported pain in their intact limb and/or lower back [11][12][13][14]. Although the weight, shape, components and alignment of the prosthesis likely affect the ability of the patient with amputations to maintain gait symmetry, in addition it is likely that a lack of confidence in the amputated limb, increased comfort by using the intact limb, and/or increased reliance on proprioceptive input from the intact limb can cause gait asymmetry [15]. During walking, the patient cannot feel the force difference between the prosthetic limb and the intact limb, and thus cannot adjust force and stance time as well as people with two intact limbs, which leads to an asymmetric gait [15]. While gait asymmetry is an important problem, few articles have addressed the possibility of reducing gait asymmetry by giving feedback to amputee patients while walking. Our hypothesis is that it is possible to train amputee patients to regain a symmetric gait pattern by offering them a real-time external feedback as a complement of proprioceptive feedback. Clinically, this training of gait pattern will be beneficial and aid in the achievement of symmetrical gait parameters, thus resulting in lower energy consumption and less complications for the sound limb and lower back.

3.2.2 Existing devices and research

Measurements of gait parameters have been performed using variety of techniques including force mats, force plates, treadmills with force plates, motion capture systems based on reflective markers and insole sensor systems.

3.2.2.1 Laboratory based systems

One widely used gait analysis tool is force mat, such as GAITRite from CIR Systems, Inc [16]. A force mat can collect data from only a limited number of steps. Treadmill systems [21][4] are also used for measuring gait parameters, but these systems are not portable. Clinical motion analysis laboratories typically contain a force plate(s) combined with a video motion capture system [17][18][19][20] using reflective markers. These systems are highly accurate but also highly expensive.

3.2.2.2 Patient mounted systems

The following devices are all patient mounted systems.

3.2.2.2.1 Foot switches. Foot switches are one of the earliest existing commercialized patient-mounted devices for gait measurement. The Portable Gait Analysis Stride Analyzer from B&L Engineering, Tustin, CA, USA [23], uses insoles with pressure switches embedded in them. Foot switches are usually low-cost with a price from less than a hundred to a few hundreds dollars (when the software and analysis equipment are included, the cost might reach to around $10k, depending on the manufacturer). But these systems are usually not suitable to measure accurate GRF patterns, which requires force sensors that provide more than just an on and off measurement. The foot-floor contact pattern measured by foot switches may also vary over different subjects, since the position of switch under the foot may change according to the size and shape of foot.

3.2.2.2.2 Insole GRF detection systems. Instead of switches, in-shoe sensor systems using different kinds of force sensors, and based on the number and accuracy of sensors embedded in one insole, the system can cost from a few hundred dollars to over $10k [24][25][26][27]. Among these, PEDAR and F-Scan are the most frequently used devices, and have been evaluated in several publications [28][29][30][31]. These systems have a high accuracy and can measure multiple gait parameters such as force distribution, stance/swing time, peak forces, etc., and can be combined with other motion capture tools such as video. But they are often prohibitively expensive for individual patient or small clinic to purchase; with cost over $10k according to different system models and software packages. These systems are designed for full analysis of foot contact force, instead of evaluating gait stance time asymmetry to provide feedback for patients with gait asymmetry.

3.2.2.2.3 Other systems under research. One of the earliest research products involving gait analysis and feedback is the leg load warning system developed by Donald Endicott [32]. Current research involving insole force sensor systems for gait evaluation include a wireless system for gait and posture analysis [33][34][35] using a 24-sensor insole pressure system based on hydrocells by Paromed [26]. This system can measure GRF from both feet at 80 Hz continuously for 5 hours, but is not specifically designed to evaluate gait asymmetry and does not give feedback to patients. Gait measurement devices that used multilayer insoles as force sensors [38][37][36] have a problem of drift on sensor readings and other devices have problems such as unsuitable for long term use due to large device size or low sampling rate [39][40]. Previous research done by Dr. Stacy Bamberg on gait shoe motion analysis instruments used multiple sensors but was not specifically designed for measuring gait asymmetry [41][42][43], and lacked feedback features.

3.2.3 LEAFS

LEAFS collects GRF data with 2 insoles containing 20 force sensing resistors (FSRs). 10 FSR signals are sampled at approximately 114 Hz, and can generate an audio feedback signal specifically for gait asymmetry in real time. At the end of each trial of walk, a summary plot of the previous walk is generated as a visual feedback, including the symmetry ratio of each step as well as overall symmetry ratio mean ($\pm$SD) and GRF curve. The signal receiver box is $1 \times 2 \times 3$ inches, and one insole with a box weighs less than 400g. It is easy to wear, and the system can be used in both indoor and outdoor environments, and can run continuously for over 10 hours on an AA battery. With the help of this system, it is now possible that low cost, long-term, easy accessible gait studies can be performed at patients' home or in a clinic that without access to a motion lab. Our goal is to use LEAFS as an assessment tool to evaluate gait stance time asymmetry in amputee population, and improve the subject's gait by offing feedback signal, to ultimately bring the gait symmetry level closer to the normal range.

3.2.4 Outline

The Methods section will discuss the LEAFS design procedure and hardware/software specifications, the evaluation method of measurement compared to a force plate, and clinical training protocols on amputee patients. The result section shows the LEAFS and force plate comparison results, and clinical test results on gait symmetry improvements of seven amputee patients over six training sessions using LEAFS. The reliability as well as the possible clinical merits of this system are discussed in the discussion section.

3.3 Methods

3.3.1 LEAFS design goals

The design goals of LEAFS are:

1. Capture heel strike and toe off, with a stance time under accuracy resolution under 10 ms;
2. Be able to be used by people with different shoe sizes;
3. Gather signal and wirelessly send to computer;
4. Provide audio and visual feedback;
5. Last over 8 hours on a single charge of batteries;
6. Be robust enough to be used in an outdoor environment.

3.3.2 System design procedure

This section is about the procedure of designing the system.

3.3.2.1 Prototype design/second version of LEAFS

The following sections will introduce the design of the second version of LEAFS.

3.3.2.1.1 Circuitry. The system prototype used Arduino Pro Mini $3.3V/8MHz$ board [44] with an atmega168 microprocessor. The Arduino Pro Mini has six analog inputs that can be read simultaneously. The Arduino board was connected to a MaxStream XBee XB24 [45] board through the serial communication port, and the XBee can wirelessly send data to a PC, or wirelessly receive a control signal from PC. Both the Arduino and XBee board were powered at 3.3V. The power supply came from a 3.7V 2000mAh polymer Lithium-Ion battery after being regulated using a PQ3RD13 voltage regulator. Based on different needs, the control board can trigger a beeper to beep, or a buzzer to buzz after receiving proper control signal from the PC; the beeper or buzzer is part of the feedback unit of the system. A small project case was used to enclose all circuitry; the case size is $1.5 \times 2.5 \times 4$ inches, shown in Figure 3.1.

3.3.2.1.2 Sensors. Force sensitive resistors (FSR, Interlink Electronics [46]) have been used in the insole sensor system. In the prototype design, there are 12 FRS, 6 sensors for each insole due to the limited number of the analog reading ports on the control board. Four FSRs were placed under the forefoot, and two FSRs were places under the hind-foot, shown in Figure 3.2.

3.3.2.1.3 Software. The software of the prototype was developed in the LabView 8.5 environment [47]. After the circuitry box reads in analog readings from the FSR and sends them to the PC, the LabView program in PC reads in signals from two USB ports (one USB port per insole) that were connected to a signal receiver shown in Figure 3.3. The software then detects heel strike and toe off timing based on the total readings from all six FSRs for each insole, and calculates stance times on both feet and the symmetry level of the subject's gait in real-time.

The software front panel is shown in the Appendix.

The symmetry level was calculated after each step using the following stance time symmetry index equation [48]:

$$SymmetryIndex(SI) = \frac{T_{Affected} - T_{Intact}}{\frac{1}{2}(T_{Affected} + T_{Intact})} \times 100\% \tag{3.1}$$

where T stands for stance time.

Based on a pre-set symmetry level threshold, the software will generate a feedback signal. When the symmetry level measured after a step falls into the threshold range, a feedback control signal will be sent to the circuitry box and triggers a beep. The base force line that used to detect a heel strike and toe off event is also present, shown in Figure 3.4 as the red line. When the total pressure reading exceeds the base force line, a heel strike event is recorded, and when the total pressure reading drops back to below the base force line, a toe off event will be recorded.

Figure 3.4 shows part of the software front panel: an example of heel strike (left) / toe off (right) event detection

3.3.2.1.4 System limitations. The prototype design read from all 12 FSR at less than 100 Hz, and needed two pairs of wireless XBee boards and occupied two USB ports on PC. The sensor layout only fit a small range of shoe sizes: US men shoe sizes 7-9. For smaller or bigger shoe sizes, the sensors can no longer capture the total ground contact force accurately. The software heel strike / toe off detection algorithm is not robust enough, since the base force line might change as the subject is walking; it is not accurate to use the same force line to calculate heel strike / toe off during the whole trial. Even though the system runs at a single charge of the battery for about 10 hours continuously, the battery used is not a regular battery type, meaning it is not easily recharged or replaced. In the improved design, these existing limitations were taken into consideration.

3.3.2.2 Improved design/third version of LEAFS

The current LEAFS consists of a hardware system using Texas Instrument®. MSP430 control chip [49], and a software system developed in MATLAB®. R2007B environment [50].

The hardware of this system, shown in Figure 3.5, consists of two wireless signal transmitters and receiver boxes that contain the MSP430 chip and one regular 1.5V AA battery and a pair of silicon insoles with 10 force-sensitive-resistors (FSR) embedded in each.

Figure 3.5 shows the improved LEAFS hardware, showing the two insoles sizes and locations of the force sensitive resistors. The signal transmitter and receiver boxes are next to the insoles

The MSP430 chip reads from nine sensors of the corresponding insole at the speed of 141 Hz and sends the data wirelessly to the software system. The extra FSR is for back up use for a shoe size bigger than US men size 10. One master receiver chip, shown in Figure 3.6, is connected to PC through one USB port.

The function of the system software is to analyze the force sensors' data, detect heel strike and toe off time of the subject's gait, calculate stance time of both feet, and calculate the Symmetry Ratios (STR) [48]:

$$SymmetryRatio(SR) = \frac{T_{Affected}}{T_{Intact}} \times 100\% \tag{3.2}$$

Based on this ratio and the feedback specifications pre-set by the physical therapist in the user interface shown in Figure 3.7, it will provide the appropriate feedback signal in real-time.

Figure 3.7 shows the user interface front panel of software system. The software records the patient's ID, prosthesis side, and trial name. The physical therapist determines the symmetry range that sets the feedback signal generation.

LEAFS has two types of feedback signals, audio and visual. The audio feedback signal is basically a beeping sound. There are three types of sounds. In negative feedback, a low beep means the patients needs to spend more time on his/her prosthetic limb and a high beep means the patients needs to spend less time on his/her prosthetic limb. In positive feedback, a nice dinging sound means the previous step was a good step. The visual feedback is generated at the end of each walking trial on PC screen. It can quickly show the summary of the previous walking trial, including the mean and standard deviation of the symmetry ratios of reach walks, and the overall symmetry ratios curves of each walks as well.

3.3.2.3 Improvements

The new system can read from 18 FSRs at 114 Hz. There are two slave chips for transmitting data, one for each insole, and one master signal receiver chip that receives data from both slave chips and communicates to the PC; thus there is only one USB port needed. The sensor layout was designed after evaluating different shoe sizes and recording the positions where the sensor readings are highest during walking. Now, this new sensor layout can fit a wider range of shoes, from US men shoe size 5 (equivalent to US women shoe size 6.5) to US men shoe size 13 (equivalent 11.69 inches). The size of the insole can be adjusted using spacers, shown in Figure 3.5, and the positions of the sensors can be adjusted as well. During a training session, one test walk was conducted at the beginning, and the sensors that are the most activated during stance phase and least activated during swing phase are selected for calculating heel strike and toe off timing in the trial. This is equivalent to modifying the layout of the sensors under the feet for different subjects and allows the sensor layout to capture the GRF well with a variety of shoe sizes. The new system uses regular AA batteries, and runs for over 10 hours continuously using two new AA batteries, one for each circuitry box. Since a regular AA battery is easy to purchase and charge, the user can just replace the batteries when needed, or rechargeable batteries can also be used.

The software automatically adjusts the base force line throughout the trial, and the algorithm for capturing heel strike / toe off was modified. It, now, not only uses base force line, but also uses the slope of the total force curve to calculate heel strike / toe off. This algorithm was proved to be more robust and able to capture heel strike / toe off more accurately, as shown in later sections of this Chapter. The modified system used Symmetry Ratio (SR), equation 3.2, instead of the Symmetry Index (SI), equation 3.1, because SR is more likely to capture an asymmetric gait than SI [48].

3.3.3 System validation test

The accuracy of the LEAFS at detecting stance time, as well as measuring SR, was evaluated through a direct comparison with force plate measurements. Five control subjects and seven subjects with amputation were recruited in this validation test.

3.3.3.1 Test procedure

The force plate measurement was conducted in the Department of Physical Therapy's Motion Capture Lab (MOCAP) [51], where two AMTI MSA-6 Biomechanics Platforms (force plates) [18] were implemented side by side. The force plate recorded data at 1000Hz or 114.068Hz, and was resampled to 114 Hz when compare with LEAFS measurement in post processing procedure.

The control subjects were asked to walk such that each foot hit one of the force plates. Normally, targeting is avoided, but the purpose was to compare simultaneous measurement. The subjects with amputations were asked to walk naturally; with the typical shorter stride length, two steps on the force plates were commonly achieved. When the subject's stride length was longer than the separation of the two force plates, only one of his/her foot can be recorded, and three records for each foot were recorded separately, as shown in Figure 3.8. Stance time was recorded simultaneously on the force plate system and LEAFS. A total of three to 10 good trials with the subject's left and right feet each captured on a force plate were recorded. Ten steps for each foot for normal gait were recorded for control subject and subjects with amputations. Additionally, control subjects were asked to walk with a limp on their left and then their right sides, 10 steps were recorded for each.

3.3.4 Clinical test procedure

Subjects were first evaluated using MOCAP equipment and LEAFS. The data records included stance time and stance time symmetry Ratio (SR) to be used as baseline data.

Next six training sessions using LEAFS were conducted over a 3 weeks' period, with each training session lasts 30 minutes. These training sessions were primarily conducted in Rehabilitation Center at University Hospital. The first four subjects received their first session immediately following the baseline test.

One week after all six sessions of training were completed, the subjects returned to MOCAP to test stance time and symmetry ratio. This test is to see if the LEAFS improves the subjects's walking ability or not, when it does improve, to investigate whether this improvement has a lasting effect or not. A subset of the subjects will be invited to return at 6 weeks for later follow-up.

3.4 Results

3.4.1 LEAFS compared to force plate

TABLE 3.1 at the of Chapter shows the subjects' basic information, and the results of the comparison are in TABLE 3.2.

In TABLE 3.2 values were calculated using original force plate measurements at 114.068 Hz minus LEAFS measurement.

The measurements for each event (limp on left, limp on right and normal walk) include left stance time, right stance and symmetry ratio measurements, which are presented in sequence in the blocks under each event.

As shown in TABLE 3.1 at the of Chapter, five control subjects were recruited, with two female and three male subjects. Shoe sizes ranged from women's 7 to men's 11.

Force plate validation results in TABLE 3.2 showed that the LEAFS left stance time measurement is 3.8±31.4 ms longer than the force plate measure on average, and right stance time measurement is 2.6±26.7 ms longer on average, with an SR 0.003±0.05 smaller than force plate measurement on average. Here three types of walking patterns were included in the calculation: left limp walk, right limp walk, and normal walks. Among three types of walks similar accuracies were shown.

TABLE 3.3 shows the information of the subjects with amputations, with three female and one male subjects, and shoe sizes ranging from women's 7.5 to men's 12.

As shown in TABLE 3.4, among LEAFS measurement on subjects with amputation, the left stance time is 3.9±27.0 ms longer than force plate measurement on average, the right stance time is 3.2±24.7 ms longer on average, with an ST 0.008±0.04 smaller on average.

Figure 3.9 and Figure 3.10 show example measurement results from control subjects No. 3 and subject with amputation No. 3. The first part of the figure showed the comparison of left stance time. The second part shows the comparison of right stance time and the third showed the comparison of SR. All these measurement were recorded during different events. For control subjects, there are three events: subject trying to make a left limp, right limp, and normal walk. On subject with amputations, the three events are their first force plate test (the one before the first training session), their second force plate walks (the one immediately after their first training session), and their third force plate walks (one week after their sixth training session).

In both Figures, the green line represents the force plate measurement at the original data rate, the red line represents the force plate measurement at the re-sampled data rate, and the blue line represents the LEAFS measurements.

Shoe size is in US size. BKA stands for below knee amputation.

In TABLE 3.4, values were calculated using original force plate measurement at 1000 Hz minus LEAFS measurement. The measurements for each event (limp on left, limp on right and normal walk) include left stance time, right stance and symmetry ratio measurements, which are presented in sequence in the blocks under each event. *NAN means that this subject did not do this test, or his/her test cannot get this type of data.

3.4.2 Clinical training results

This section shows the measurement of symmetry ratio using LEAFS on both control subjects and subjects with amputations.

Figure 3.11 and Figure 3.12, show the subjects' normal walks in a hallway inside a building wearing LEAFS. The black line represents the mean and standard deviation of ratio of the walks. As shown in the figures, the control subject's ratio is generally between 0.9 and 1.1; the subject with amputation's ratio is generally below 0.9 (these data came from the subject's first training session).

TABLE 3.5 and TABLE 3.6 show the normal walk ratio ranges of control subjects and subjects with amputations. Compared to previous study results [52], mean and SD of stance time ratio of healthy population (n=81) is 1.02±0.02, this study showed similar results for individual subjects. All five control subjects are the same control subjects from the force plate validation tests, and the walking trials were conducted following the force plate validation trials.

Normal walks means that the subjects were walking at their self-selected speed, wearing their own shoes with LEAFS insoles in them, in a hall way inside of a building, and with no or very few distraction factors such as noise.

Here the subjects' normal walks used their first training sessions' data.

The ratio for subjects with amputations used stance time on prosthetic side/stance time on intact side.

The subject with ID number 2 has been using a prosthesis for 9.5 years, and showed little gait asymmetry in hallway walk tests. Some subjects have been using a prosthesis for less than 3 years.

3.4.2.1 Clinical test all training sessions' results

These results show each subject's changes over all six (or five) training sessions.

TABLE 3.7 shows the mean and SD of subjects' first and last training sessions. The clinical significance was determined based on improvement between these two sessions.

The fifth training session of the subject with amputations No. 1 is shown in Figure 3.13. The magenta line represents the lower threshold for feedback signal. The sixth training session result of subject with amputation No. 4 is shown in Figure 3.14. The red line in Figure 3.14 represents the higher threshold for feedback.

3.5 Discussion

3.5.1 System accuracy compared to force plate

Overall, LEAFS showed reasonable accuracy in measurements on both control subjects and subjects with amputations. Both groups of subjects covered a reasonable wide range of shoe sizes (women's 7 to men's 12). This showed the adjustability of the LEAFS insoles.

The mean value of LEAFS measurement stance time on control subjects is generally overlapping the mean value of the force plate measurements (overall difference less than 15ms), but the standard deviation is high (overall SD of difference is within $\pm$32ms). But the overall standard deviation of mean of difference is low, which means LEAFS can be used to measure the mean stance time of a trial of walk that contains multiple steps (more than 10 steps), and maintain a reasonable accuracy, but cannot be used to measure the stance time of a single step.

Although LEAFS has an acceptable accuracy of measuring mean stance and mean symmetry ratio over a trial of steps, it is not accurate at measuring stance time and symmetry ratio of a single step.

3.5.2 Error source of measurement

LEAFS has a data rate of 114 Hz; thus its maximum resolution of stance time measurement is about 8.8 ms. The mean and standard deviation of the error of LEAFS measure is within 4$\pm$32ms, which is about three times of the system resolution. To decrease this error, a first choice is to increase the system resolution by increasing the data rate of LEAFS. It has been shown by experiments that a data rate of 250Hz is achievable. It is possible that if the system runs at 250Hz, which means the resolution of measurement is 4ms, then the error will be decreased to three times of the resolution, which is 12ms.

Another source of error is the algorithm of detecting heel strike and toe off. Although it seems easy for the human eye to identify a heel strike and toe off on the ground contact force curve, it is complicated to detect these two events consistently and accurately using the same algorithm, particularly since the algorithm must be able to operate in real-time. Different people have different ground contact force curve signature, and factors such as type of shoe they are wearing, their health condition, whether they are fatigued or not, etc. can affect their force signature. The algorithm must be robust and flexible at adapting to different walking patterns, and thus, more factors should be considered beyond the trigger level and slope that are currently used.

As implied in the methods section of this paper, the sensor locations are extremely important, because they are directly related to what kind of ground contact force curve signature we might see. An investigation of the data of the control subjects and subjects with amputations showed that different individuals' force signatures are better represented by a different selection of sensors among the existing 10 sensors. For example, individuals with larger feet benefit from using the most anterior sensor, because it is closer to the great toe, while individuals with medium or small size of feet benefit from using the second-most anterior sensor. This situation applies to the rest of the sensors. While there are a few choices for the sensors under the metatarsals, the heel sensors stays the same for all sizes of feet. It is possible that, with more sensors in the insole, more appropriate sensor locations could be available for different sizes of feet, and that this would improve the measurement accuracy, through more consistent force signatures. This will not only contribute to improved stance time and symmetry ratio measurements, but also to the center of pressure curve measurement.

A clear location strategy has not been established. Further study will be discussed in future work.

3.5.3 Symmetry ratio ranges for different population

As shown in [48], a healthy population with n=81 has a symmetry ratio value of 1.02±0.02 (in this calculation, all ratios that are less than 1.0 have been replaced by its inverse). The LEAFS measurement showed a similar result. Based on a group of five people, the mean and SD of symmetry ratio is 1.01±0.03.

In the group of subjects with amputations, the mean symmetry ratio varies depending on how long the subject has been wearing the prosthesis. Usually the longer the subject has been using a prosthetic leg, the lower asymmetry in his/her gait is shown. Subject No. 2 with 9.5 years of amputation history, showed little asymmetry according to her mean symmetry ratio, but the corresponding standard deviation (0.081) is higher than control subjects (0.035). This indicates that her gait has more variability than control subjects, possibly caused by wearing the prosthesis. In subjects with less experience using prosthesis, more asymmetry is evident, and in subject No. 4, who has only been using prosthesis for 7 months, the most asymmetry was seen. However, some researchers have demonstrated that asymmetry in gait measures, such as stance/swing ratio and period of double support, is not significant among patients with 3 years experience using prosthetic limb [15], while other researchers demonstrated that patients with over 10 years experience using prosthetic limb still have gait asymmetry [8].

Based on the symmetry ratios, the LEAFS might be more helpful to people who have less experience using prosthesis, since they have more potential to improve their gait symmetry level. The earlier they can learn to walk symmetrically, the less future damage there will be to their intact limbs.

3.5.4 Clinical training effectiveness

Of the seven subjects, four showed improvement on their gait asymmetry. One maintained her symmetry level at the normal level (she started with no gait asymmetry). In each session, the first trial is not considered, since that trial was used to determine where to set the threshold. The total number of trials in each session depended on the subjects' ability of walking. If the subjects showed sign of fatigue, the trial (or session as necessary) was stopped. Shown in Figure 3.13 is an example of one session of all six training sessions for a single subject. Visible improvement can be seen, with an overall trend that is visibly increasing, indicating it is possible for LEAFS to have a short time period effect on improving gait.

Between sessions, even though the subject is not wearing LEAFS, it is possible that he/she remembers the feedback signal and the implying message, typically that a longer step on their prosthesis is required. Thus, it is possible that some improvement will also be made during the time period between sessions. As shown in Figure 3.14, there were visible improvements between sessions. In the case of subject No. 4, this trend is more obvious because he is a relatively new user of a prosthetic leg, and has a strong personal will to learn to walk symmetrically whenever he walks. Thus, LEAFS has the potential to influence the daily walking pattern of the subjects.

Over the six sessions, the subject's gait symmetry ratio gets closer to 1, which means perfectly equal on both sides. Amputee gait often has a higher standard deviation, because it is usually hard to maintain balance on a prosthesis, as shown in TABLE 3.7. Ratio stands for symmetry ratio, ratio = stance time on prosthetic side / stance time on intact side. However, based on the mean symmetry ratio level of each session, improvement is evident.

3.5.5 Individual results

We will discuss individual results in the following paragraphs.

3.5.5.1 Trigger level adjustment

There are two types of feedback signal, a negative feedback and a positive feedback. When working on the negative feedback mode, the patient will hear a beep when his/her gait symmetry level falls out of the range of the upper and lower trigger level, so that he/she can focus on what a bad step is; when working on the positive feedback mode, the patient will hear a different dinging sound when his/her step ratio falls into the range of upper and lower trigger level, so that he/she can focus on what a good step is (silence follows a step outside of the trigger level). The range between lower threshold and higher threshold of negative feedback is often larger than that of the positive feedback.

In Figure 3.13, all trials are using negative feedback, and the subject's gait symmetry level is mainly below 1; thus it is the lower trigger level that determines whether the feedback signal should be given out or not. When the subject's gait improved to a level that very few steps have a symmetry level below the trigger level, the trigger level was raised. It is important not to set the lower trigger too high, but rather to choose an appropriate level such that the patient can be challenged and yet not too frustrated by the beeping feedback.

In Figure 3.14, the fifth training session mainly used the positive feedback, the patient came to a point where a bad step is slightly below 0.9, and there is little difference compared to a good step that is above 0.9. Thus the trigger level is set to be above 0.9, and whenever the patient made a good step of ratio above the trigger, an encouraging dinging feedback sound was given. In this process, the patient gradually leaned what a good step is and focus more on making a good step other than correcting a wrong one.

3.5.5.2 Improvement pattern over 6 sessions

All subjects who have an increase in their SR showed a similar pattern in their training sessions. As shown in Figure 3.14, at the beginning of training, the subject tried to adjust their gait and increase their stance time on prosthetic side. In this period, negative feedback is often used, and it is the lower trigger level determines the feedback signal. While the subjects are trying change of gait, ratio of that standard deviation increases. Then, when they reach to a point where their symmetry ratio is close to 1.0, and they are still trying to increase their stance time on prosthetic side, they start to over-compensate. Then the upper trigger level plays a more important role. Whenever they limped on the other side, a warning sound will be given out. At the last stage of training sessions, the subjects learned to walk more towards the perfect ratio, and it is often hard for them to take the exact symmetry step, so their STR standard deviation reaches the peak. Then, the positive feedback will be used instead of the negative feedback, so that the patient can focus more on his/her good steps. Then if the training session is long enough, the subject will gradually learn to walk symmetrically without over-compensating, and the STR standard deviation will decrease and drops back to normal level.

Some of the subjects showed only the first two or first three stages. Only two subjects showed the whole four stages. It is likely that more training was necessary for the other subjects.

3.5.6 Short-term usage disadvantages

As discussed in the previous section, not all subjects showed the full four stages of improvement, because there were not enough training sessions for them to actually achieve to that level. Given that each subject can have more than six training sessions, it is highly possible that more subjects will experience all four stages of improvement and finally learned to walk with out any stance time asymmetry.

Also, for one to completely adapt to a new walking pattern, training sessions in hospital might not be enough; instead, training at home would be more effective. In that case they can monitor their gait while they are doing daily walks. Thus once they learned to walk symmetrically, it is less likely for them to forget and start walking asymmetrically again.

3.6 Future studies

3.6.1 Long-term daily usage VS short-term clinical usage

Our future research will extend the training sessions to time period longer than 3 weeks, and will also send the device to the patients' homes so that training can be continuous and take place in daily activities. The long-term home usage of LEAFS will be investigated as compared to the short-term clinic usage investigated in this study.

3.6.2 Improvement on system accuracy

This study showed that the accuracy of stance time and symmetry ratio measurement still needs to be improved. The relationship of accuracy and sensor locations will be investigated. The algorithm of detecting heel strike and toe off in real time will also be improved.

3.7 Conclusion

Overall, LEAFS showed reasonable accuracy in stance time measurement and symmetry ratio measurement as compared to a force plate measurement. The system showed better accuracy at measuring mean stance time and mean symmetry ratio over a trial of walk that contains multiple steps (more than 10).

A clinical gait training test showed that LEAFS has the potential ability to train people with amputations to regain a symmetrical gait.

3.8 Acknowledgements

The authors acknowledge the support from University Hospital Rehabilitation Clinic, as well as the patient participants for this study, for without their help this study cannot be completed. Also, thanks for the support from Motion Capture Lab from University of Utah Department of Physical Therapy. Thanks to all the participating researchers in this study.

Figure 3.1. Circuitry box in the prototype design.

Figure 3.2. Insole sensors in the prototype design.

Table 3.1. Control subjects' general information

ID	Age	Gender(M/F)	Height(ft/inch)	Weight(Lbs)	Shoe Size(US Size)
1	23	M	$5'10''$	145	M 10.5
2	22	F	$5'5''$	100	W 7
3	27	M	$5'7''$	150	M 9.5
4	38	M	$5'11''$	178	M 11
5	35	F	$5'6''$	150	W 8.5

Table 3.2. LEAFS compare with force plate on control subjects

ID	Note	Left Limp	Right Limp	Normal walk	Overall
1	L stance(ms)	2.9±10.2	1.9±17.6	-23.3±55.9	-6.1±35.4
	R stance(ms)	-9.1±7.3	-10.6±7.6	-16.4±13.0	-12.0±9.9
	SR	0.01±0.02	0.02±0.03	-0.009±0.09	0.008±0.06
2	L stance(ms)	-11.5±64.8	14.2±13.2	-0.1±10.7	0.9±38.8
	R stance(ms)	-29.5±20.6	-11.4±11.2	0.3±7.9	-13.5±18.6
	SR	0.02±0.07	0.06±0.05	-0.001±0.01	0.03±0.06
3	L stance(ms)	3.0±11.6	-15.7±9.7	9.9±7.9	-0.9±14.5
	R stance(ms)	-5.8±11.8	-19.9±7.8	-5.8±5.8	-10.5±10.9
	SR	0.005±0.01	0.040±0.04	0.02±0.01	0.02±0.03
4	L stance(ms)	-0.6±19.6	-6.0±8.3	8.2±25.1	0.5±19.3
	R stance(ms)	9.8±27.4	19.5±23.3	24.5±8.5	17.9±21.5
	SR	-0.01±0.03	-0.04±0.05	-0.02±0.03	-0.02±0.04
5	L stance(ms)	16.6±49.3	-44.0±16.2	-12.9±18.7	-13.4±39.7
	R stance(ms)	34.2±48.5	-23.0±39.0	3.7±18.4	5.0±43.3
	SR	-0.01±0.07	-0.02±0.09	-0.02±0.04	-0.02±0.07
Overall		2.1±37.6	-9.9±23.7	-3.6±30.9	-3.8±31.4
		-0.1±33.8	-9.1±25.6	1.3±17.6	-2.6±26.7
		0.001±0.05	0.01±0.07	-0.007±0.05	0.003±0.05

Table 3.3. Subjects with amputations' general information

No.	Age	M/F	Height (ft/inch)	Weight (Lbs)	Shoe (US)	Amp. side	Time since	Amp. level	Amp. reason
1	62	F	5'2"	127	W 7. 5	L	1. 5 y	BKA	Unknown
2	50	F	5'9"	201	W 9.5	L	9. 5 y	BKA	Unknown
3	57	F	5'3"	259	W 8	L	2. 5 y	BKA	Unknown
4	65	M	6'4"	256	M 12	L	7 m	BKA	Injury
5	61	M	6'1"	263	M 10	R	2. 5 y	BKA	Unknown
6	22	F	5'3"	140	W 7. 5	L	5. 5 y	BKA	Injury
7	62	F	5'7"	149	W 9	R	30 y	BKA	Unknown

Figure 3.3. XBee wireless signal transmitter.

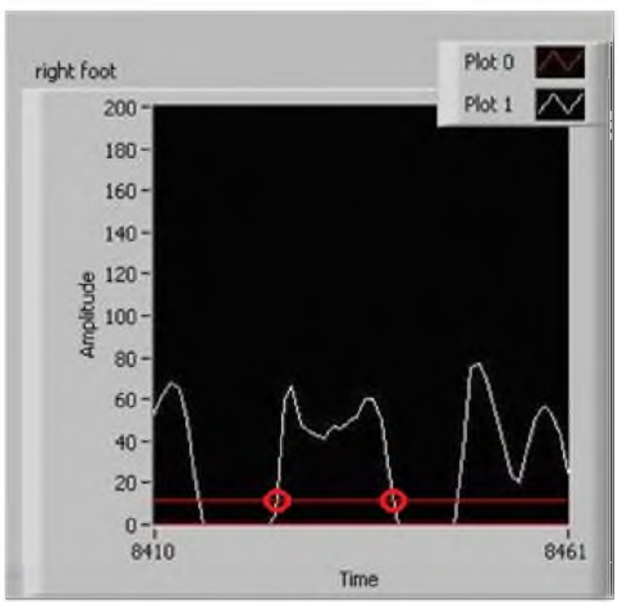

Figure 3.4. Part of the software front panel.

Figure 3.5. TI wireless signal transmitter.

Figure 3.6. Signal receiver.

Figure 3.7. The user interface front panel.

Figure 3.8. Measurement on force plate.

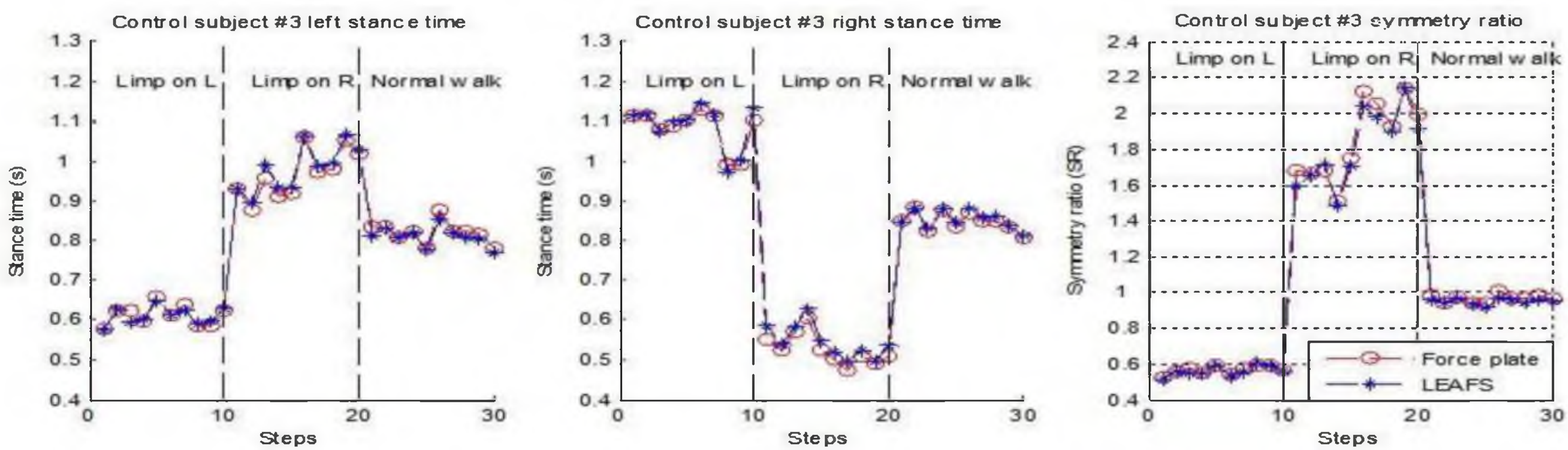

Figure 3.9. Control subject No. 3 force plate walks

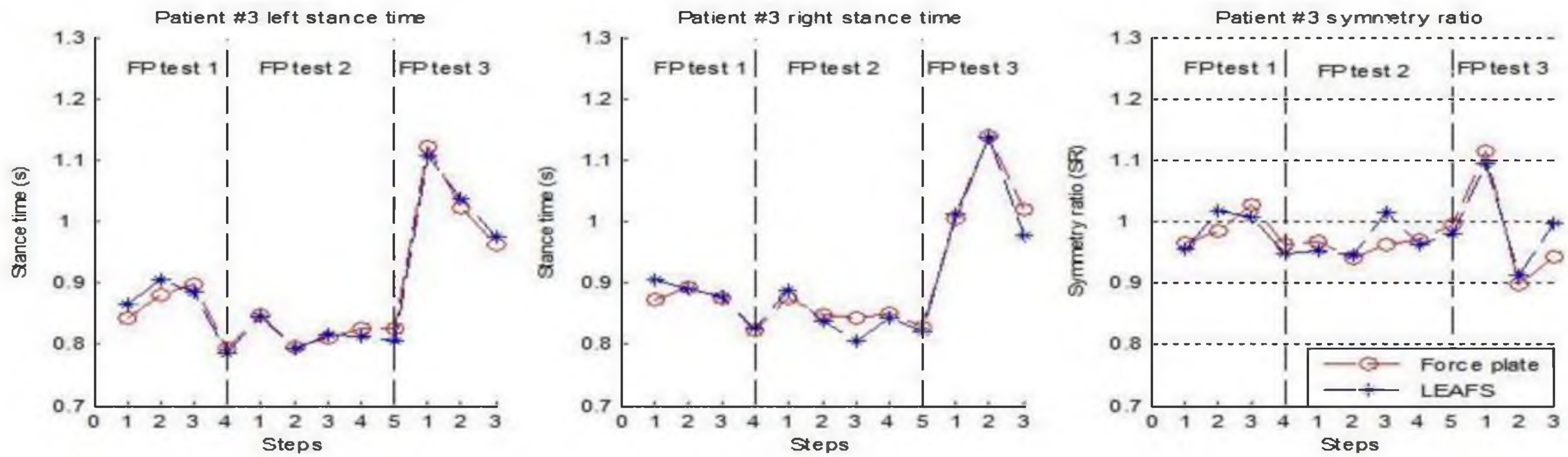

Figure 3.10. Subject with amputation No. 3 force plate walks

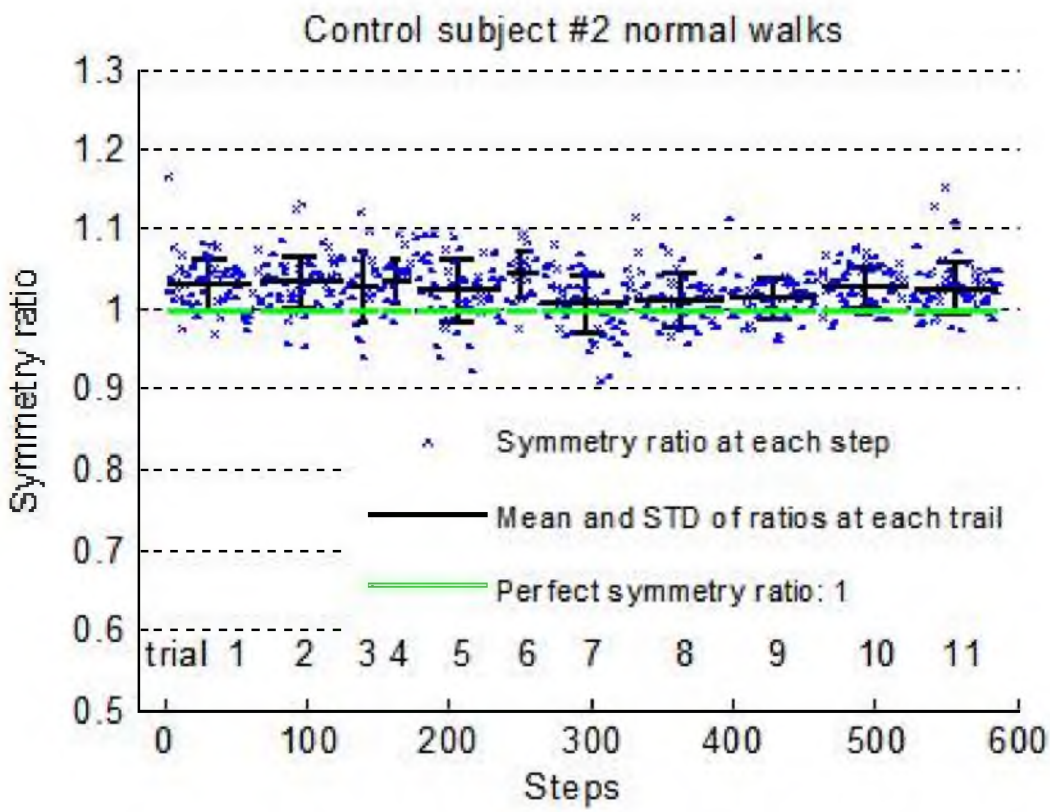

Figure 3.11. Control subject No. 2, normal walks

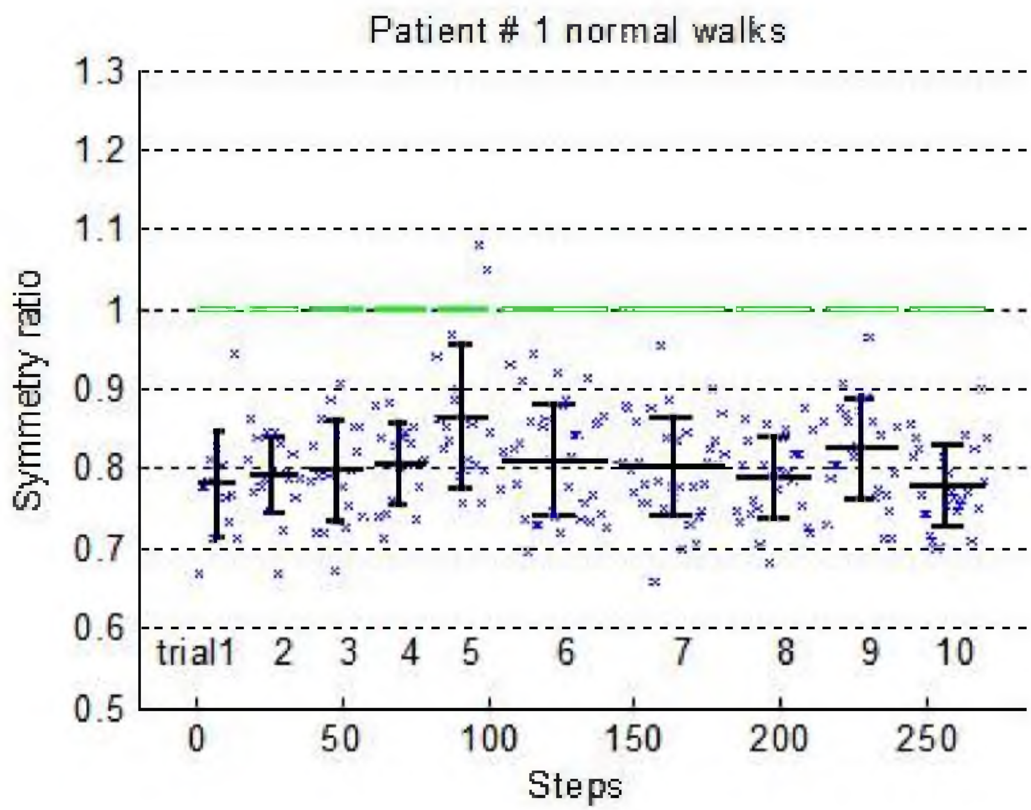

Figure 3.12. Subject with amputation No. 2, normal walks

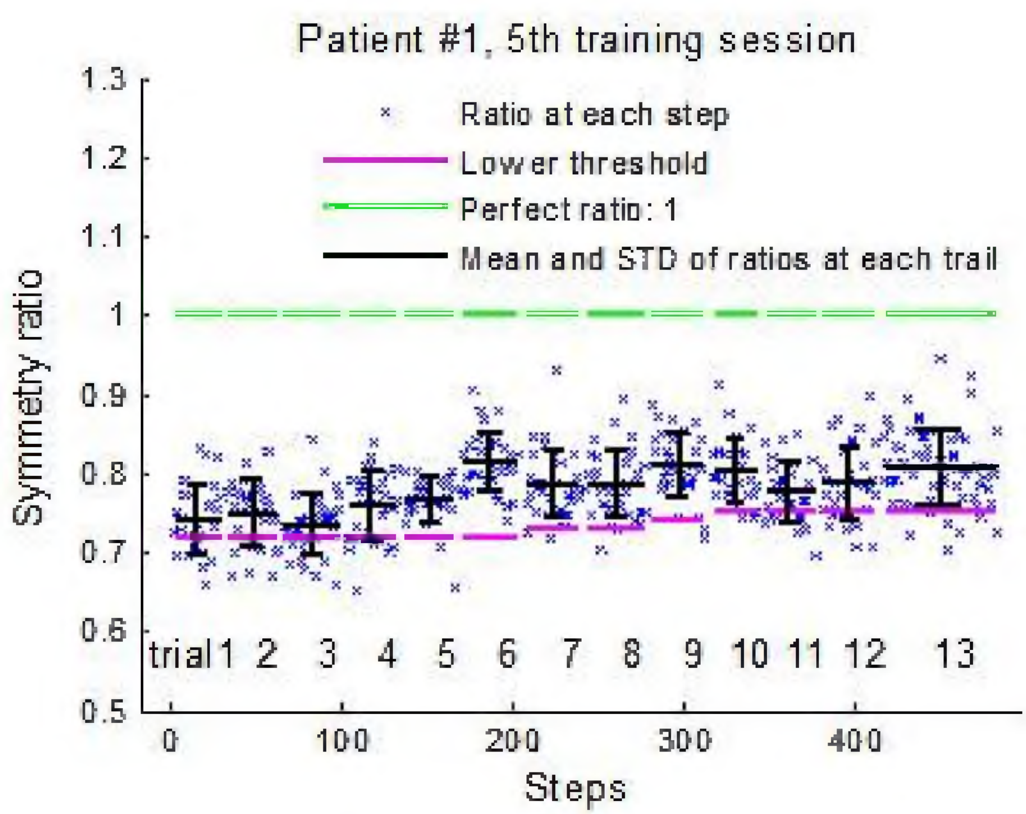

Figure 3.13. Subject with amputation No. 1, the fifth training session

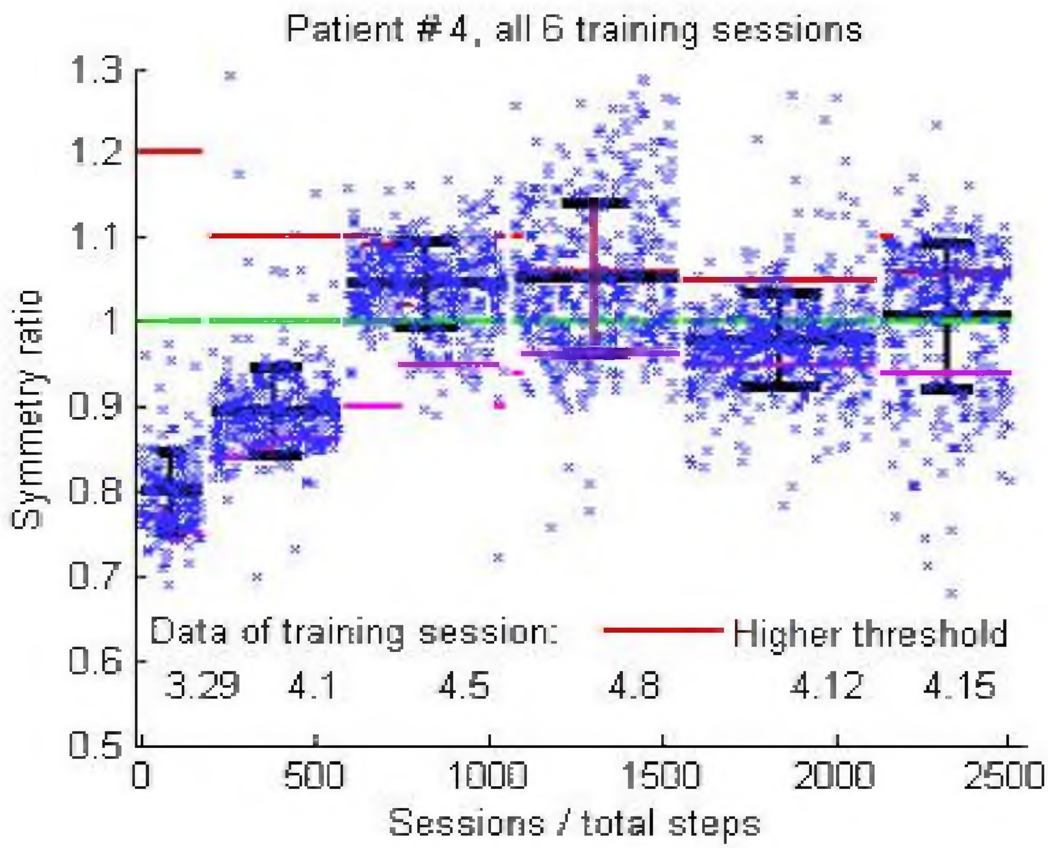

Figure 3.14. Subject with amputation No. 4, all six training sessions

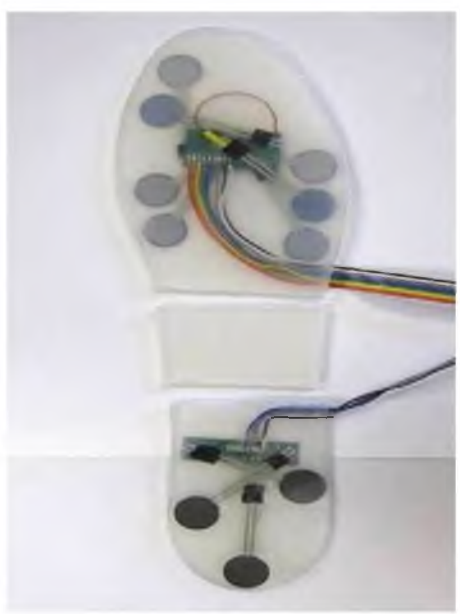

Figure 3.15. Sensor locations

Table 3.4. LEAFS VS Force plate on subjects with amputations

ID No.	Note	1st FP test	2nd FP test	3rd FP test	Overall
1	L stance(ms)	21.4±15.0	-3.4±35.1	16.8±12.8	12.0±23.2
	R stance(ms)	19.1±29.6	-18.3±6.9	-0.7±11.8	-1.3±22.3
	ratio	0.006±0.03	0.01±0.04	0.02±0.03	0.01±0.03
2	L stance(ms)	-6.9±4.9	-7.4±4.9	*NAN	-7.2±4.6
	R stance(ms)	16.7±7.2	-4.8±17.3		4.8±17.3
	ratio	-0.008±0.00	-0.003±0.02		-0.004±0.02
3	L stance(ms)	-6.4±20.3	6.6±9.9	-3.5±16.6	-0.3±15.4
	R stance(ms)	-8.8±16.2	10.2±18.0	13.6±25.7	4.7±20.3
	ratio	0.003±0.02	-0.004±0.03	-0.02±0.04	-0.005±0.03
4	L stance(ms)	-15.5±8.8	-1.2±6.4	-25.2±27.2	-14.0±18.0
	R stance(ms)	-16.7±9.4	-2.5±3.8	8.9±4.6	-5.9±13.0
	ratio	NAN	NAN	NAN	NAN
5	L stance(ms)	-21.5±8.5	NAN	6.2±13.2	-7.6±18.1
	R stance(ms)	-16.6±5.0	NAN	-33.2±1.4	-24.9±9.7
	ratio	NAN	NAN	NAN	-0.05±0.00
6	L stance(ms)	-15.9±3.3	NAN	31.2±42.2	4.3±35.1
	R stance(ms)	-0.8±27.8	NAN	-6.8±13.1	-3.4±21.3
	ratio	-0.02±0.02	NAN	0.05±0.05	0.01±0.05
7	L stance(ms)	0.9±4.2	NAN	-82.8±25.1	-41.0±48.6
	R stance(ms)	-6.2±8.9	NAN	-32.4±79.7	-19.3±52.7
	ratio	-0.009±0.008	NAN	0.044±0.065	0.017±0.050
Overall		-5.6±16.6	-1.2±17.3	-6.9±42.2	-3.9±27.0
		-1.8±21.1	-4.0±16.7	-7.7±33.4	-3.2±24.7
		-0.004±0.02	0.00±0.03	0.02±0.05	0.008±0.04

Table 3.5. Control subjects' normal walks

ID	Total trial Num	Total step num	Shoe (US)	Overall ratio (mean ± SD)	Ratio equation
1	3	200	M 10. 5	1.001 ± 0.05	L/R
2	11	600	W 7	1.01 ± 0.03	L/R
3	3	80	M 9. 5	0.97 ± 0.02	L/R
4	NAN	NAN	M 11	NAN	L/R
5	1	110	W 8	1.06 ± 0.040	L/R

Table 3.6. Subjects with amputations' normal walks

ID	Total trial num	Total step num	Shoe (US)	Overall ratio (mean ± SD)	Amp Side
1	10	270	W 7. 5	0. 80 ± 0. 07	L
2	7	280	W 9.5	1. 04 ± 0. 07	L
3	5	240	W 8	0. 94 ± 0. 06	L
4	6	190	M 12	0. 80 ± 0. 05	L
5	9	420	M 10	0.98 ± 0.09	R
6	9	700	W 7. 5	0.80 ± 0.06	L
7	11	300	W 9	0.92 ± 0.07	R

Table 3.7. Subjects with amputations' training sessions

ID	Total session num	First session (*Ratio Mean ± SD)	Last Session (Ratio Mean ± SD)	Ratio Significance
1	6	0. 80 ± 0. 07	0. 83 ± 0. 05	NO
2	6	1. 05 ± 0. 07	0. 99 ± 0. 06	NO
3	5	0. 94 ± 0. 06	1. 01 ± 0. 12	NO
4	6	0. 80 ± 0. 05	1. 006 ± 0. 09	YES
5	6	0.98 ± 0.09	0.94 ± 0.05	NO
6	6	0.80 ± 0.06	0.96 ± 0.08	YES
7	6	0.92 ± 0.07	0.90 ± 0.05	NO

APPENDIX

ADDITIONAL FIGURES

Figure A.1 Figure A.2 Figure A.3 LabView software user interface front panel

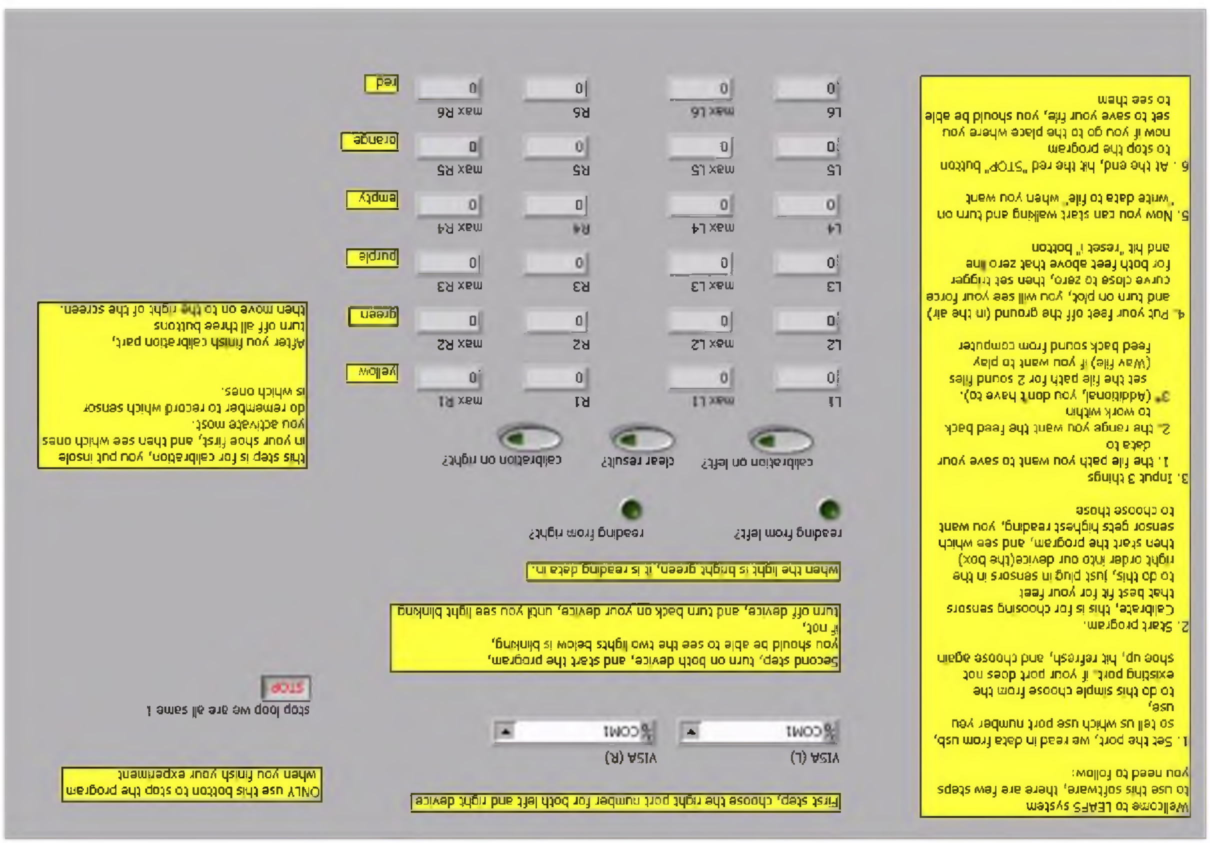

Figure A.1. LabView software user interface front panel part1

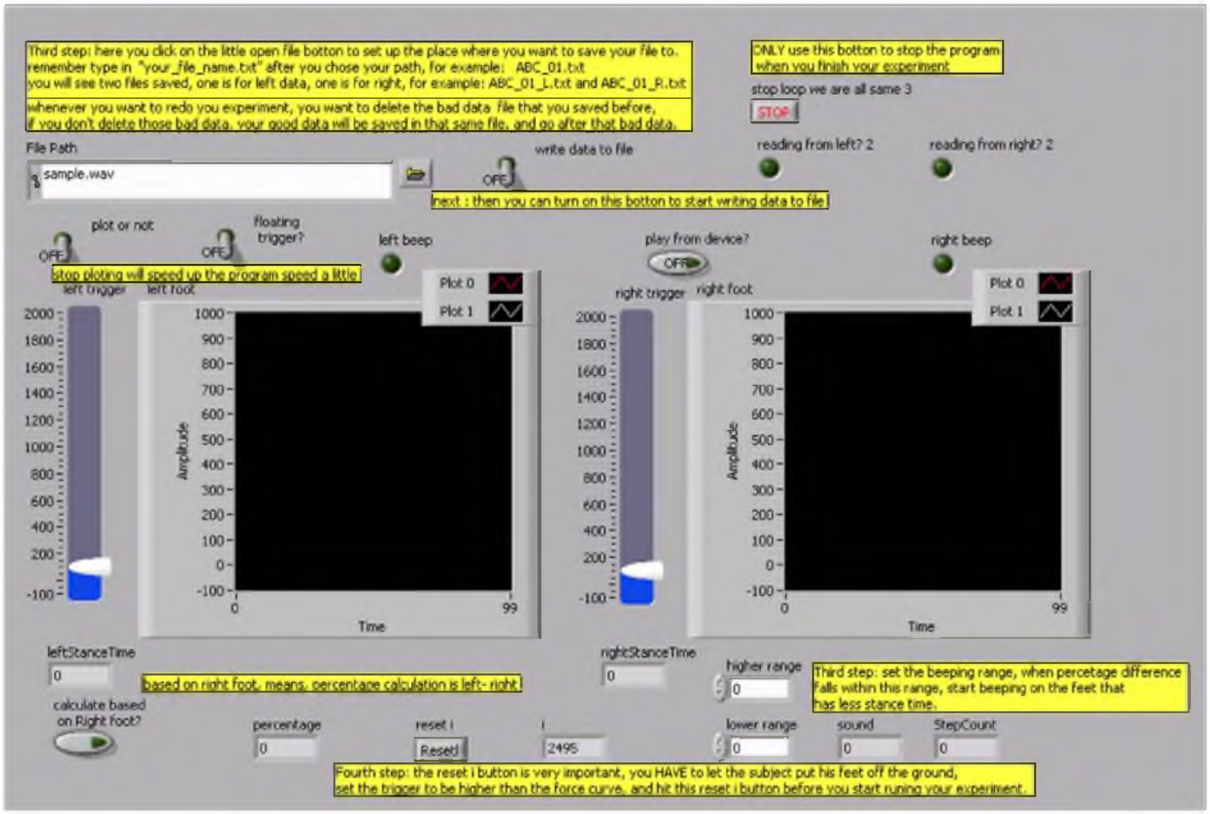

Figure A.2. LabView software user interface front panel part2

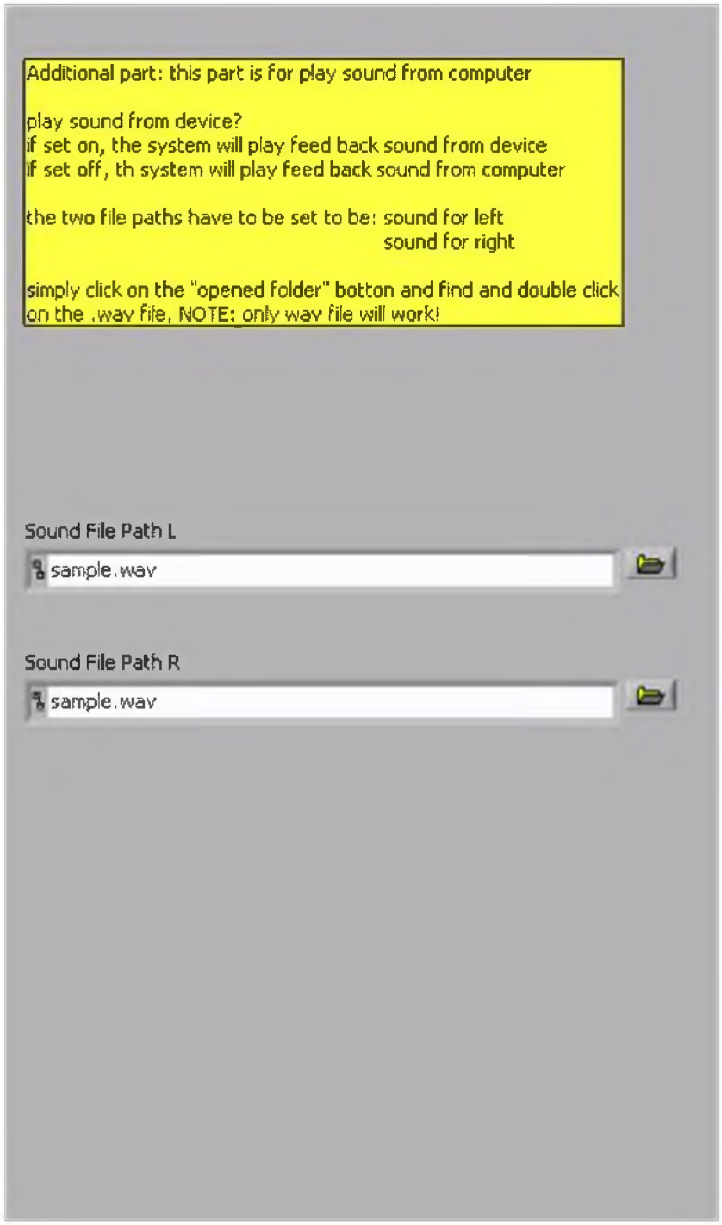

Figure A.3. LabView software user interface front panel part3

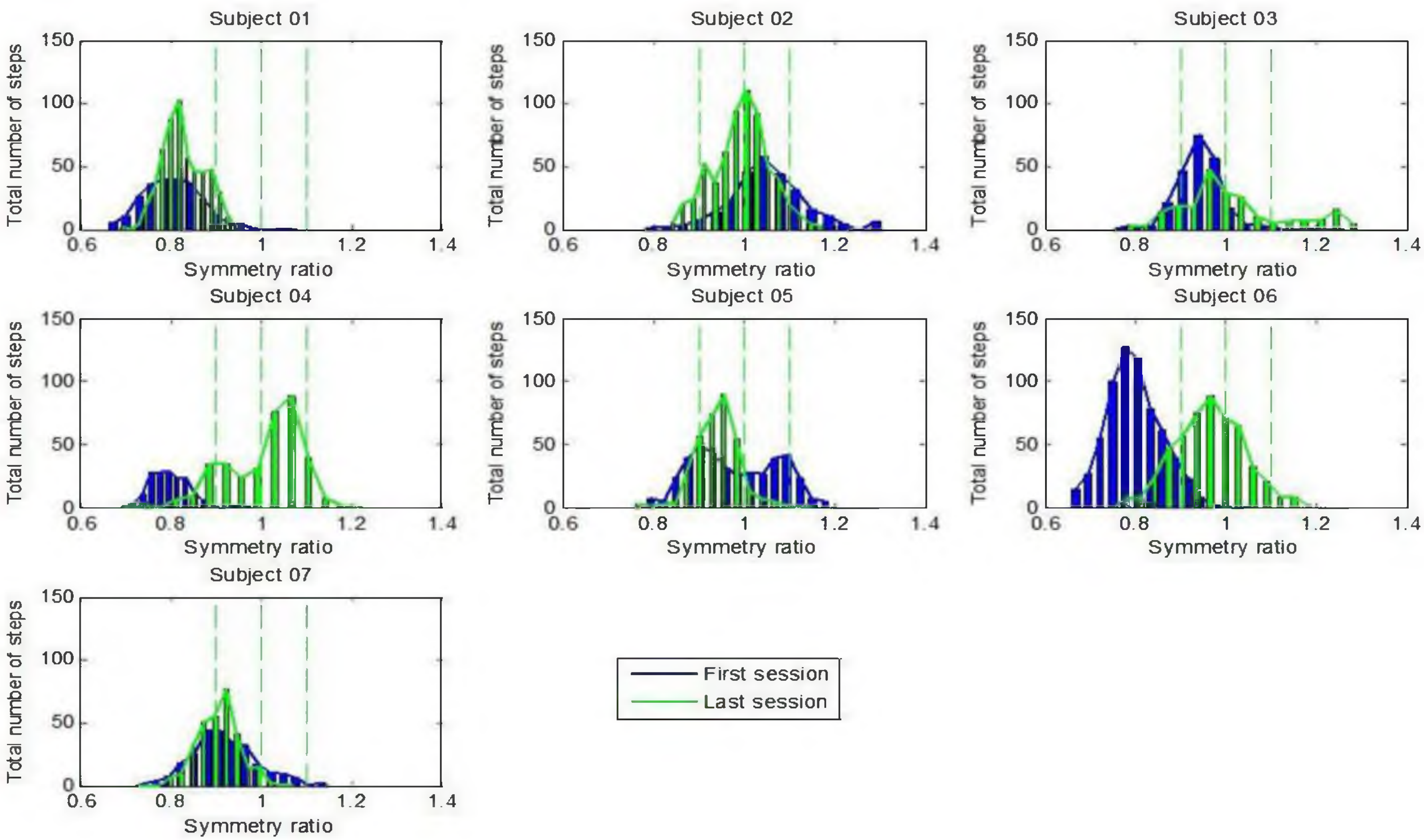

Figure A.4. Histogram showing subjects` overall improvements

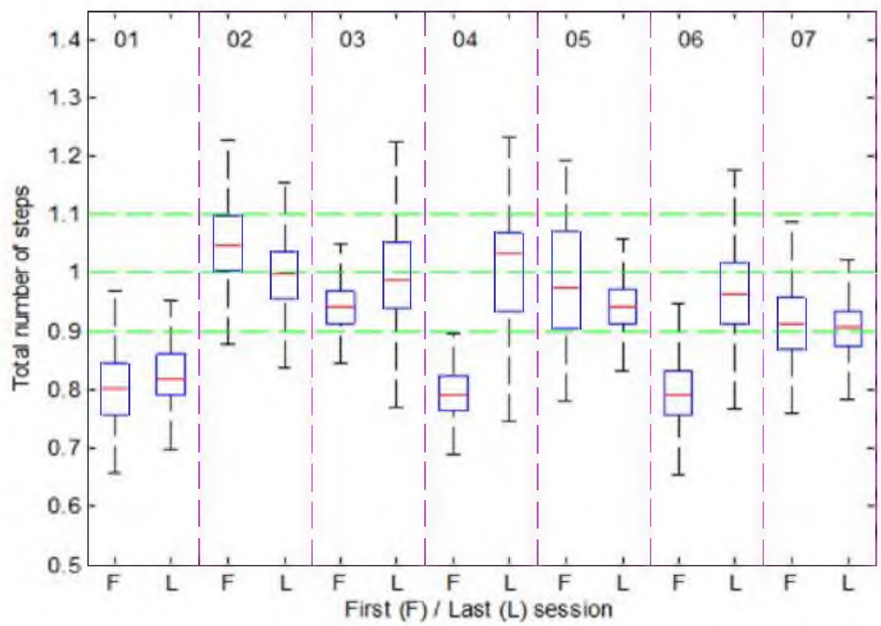

Figure A.5. Subjects' changes between first and last training sessions

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