DEVELOPMENT OF AN ALTERNATIVE SYSTEM FOR SUSPENDED GAIT ANALYSIS

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Abstract: Spinal Cord Injury (SCI) may impair an individual's gait. For these cases, a rehabilitation technique that has become more popular is Functional Electrical Stimulation (FES). Gait analysis is an important technique to evaluate rehabilitation of patients undergoing FES-assisted therapy. This work proposes a system that monitors gait variables – knee joint angles, and ground reaction forces (heel and metatarsal) – and uses them as inputs for gait analysis of paraplegic patients. The methods for building the data acquisition hardware (transducers and interface) and software are described, along with the transducer calibration methods. The results show the final prototype for the gait analysis system, which allows comparison between different individuals' gaits, as well as different rehabilitation stages for the same individual. The software has a recording feature, as well as digital control outputs, which may be used in the future for training an Artificial Neural Network (ANN) and controlling the individual's FES stimulator. In the near future, the system may be of great applicability for suspended FES-assisted gait analysis and control.

1 INTRODUCTION

1.1 Human Gait

Gait may be defined as a form of biped progression in which lower limb repetitive movements include periods of double support – in which both feet are in contact with the ground – followed by periods in which only one foot supports the body (stance) and the other is being moved above the ground (swing) (Wall, 1999).

On a normal gait, the stance phase constitutes 60% of the gait cycle, and is defined as the interval in which the reference foot is in contact with the ground. The swing phase begins with heel contact and ends when the foot leaves the ground (toe off surface).

1.2 Gait Analysis

For individuals that suffer Spinal Cord Injury (SCI), a technique that has contributed for rehabilitation is Functional Electrical Stimulation (FES) (Castro and Cliquet Jr., 2000). FES treatment may be associated with dynamic suspension (Field-Fote, 2001). This suspension allows a weight reduction, maintaining the load on the lower limbs at a level they are able to stand. It also stabilizes the trunk, resulting in better balance for the patient, and lowers upper-limb overload, frequently observed on walker-aided training for paraplegic patients.

Gait analysis is an important tool for biomechanical studies of the rehabilitation process. Veltink, Liedtke, Droog, and van der Kooij (2005) developed a gait analysis system using two commercial six-degrees-of-freedom force and moment sensors under a sandal. Giacomozzi and Macellari (1997) constructed a compound instrument by superimposing a dedicated pressure platform on a commercial force platform.

The objective of this work is the development of an alternative system for gait analysis, to be used on the evaluation of patients' rehabilitation for suspended FES-assisted gait.

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2 METHODS

2.1 Determination of Variables

The first task was to determine which variables would serve as inputs for the system. Various works have been done proposing different models (Tong and Granat, 1999; Pappas et al., 2004; Popovic et al., 1998), each one showing advantages and disadvantages. For this work, the idea was to be able to measure the actual values for ground reaction forces, and use them for comparison on gait analysis, not as triggers for detection of foot contact.

Popovic et al. (1998) stated that ground reaction forces alone could not be used for satisfactory gait phase characterizing. In order to avoid this problem, an extra variable was chosen to serve as system input. The data posted on the CGA Normative Gait Database (Kirtley, 2006) show kinematic and kinetic analysis for healthy adults and children. Taking in consideration the movement amplitude, easiness to mount sensors, sensor stability during movement and amount of signal noise, the variable selected was knee flexion/extension.

2.2 Transducer Selection

Based on the studies conducted by Cunha (1999), the chosen transducers for knee flexion/extension angle determination were shape sensors. These sensors use 0.25 mm diameter fiber optics, specially treated to lose light by refraction proportionally to the deflection suffered by the fibers. They present some important characteristics, such as light weight and a simple electronic signal processing package, incorporated to the fiber optics.

For the ground reaction forces, the analysis was based on the studies conducted by Leite (2003). The author developed an instrumented crutch to measure vertical reaction forces applied by patients during gait, using strain gages.

Strain gages are resistors composed of a very thin conductive layer over an isolating compound. The sensor is glued on a structure, and when there is some deformation caused by applied forces on the structure, it is possible to determine the value of the force, since it depends only on the type of material and the geometry of the structure.

The fact that strain gages require a rigid structure to operate represents a disadvantage for this kind of sensors. The solution found for the problem was to construct cylindrical aluminum rings to work as load cells for the strain gages. For each load cell, a metal base and a semi-spherical top were constructed, so that the top's radius helps bring the applied loads as close as possible to vertical. Figure 1 shows one of the load cells with the strain gages attached (left) and with the protective covering, metal base and top (right).

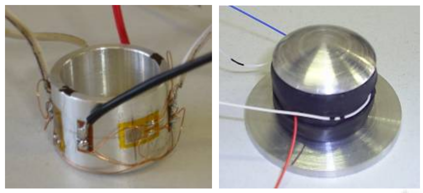

Figure 1: Ground reaction force sensors (load cells).

Three load cells were made for each foot, and they were attached to a sandal, mounted on cylindrical metal plates. One load cell was positioned in the heel area, and the other two were positioned in the metatarsal area, connected in parallel – in order to compensate different styles of stepping patterns.

2.3 Hardware Interface

The hardware interface circuit consists of anti-alias filters (cutoff frequency of 18 Hz), voltage regulators (± 5VDC), and amplifiers with adjustable gains for the strain gages. The information was acquired using a data acquisition board.

The circuit is enclosed in a box, connected to the data acquisition board, and also to a smaller interface box, which is positioned around the patient's waist, and has connection hubs for the instrumented sandals and shape sensors. The transducers connected to the patient's interface box are shown in Figure 2.

Figure 2: Transducers and patient's interface.

2.4 Transducer Calibration

To calibrate the shape sensors, they were attached to two articulated flat metal bars, and a protractor. The voltage values corresponding to each 5º angle interval, from -140º to 140º, were collected.

For the load cells, the calibration was performed using a dynamometer, and the voltage values for each 50 N interval, from 0 to 1000 N were collected. For both cases, the results showed a linear behavior.

2.5 Software Interface

The software used to program the system was LabVIEW 6.1. It has some advantages, including user-friendly graphic interface, and compatibility with the data acquisition board.

The interface software has a monitoring module, which allows visually following the behavior of knee flexion/extension angles, and ground reaction forces – heel and metatarsal. It is also possible to record them in a spreadsheet file for data analysis.
The visual interface shows three screens (selectable using tabs): the "Main" screen, which has all the controls and configuration options; the "Angles" screen, which shows the graphics for the knee flexion/extension angles, and the "Forces" screen, which shows the graphics for the ground reaction forces. Figure 3 illustrates the "Forces" screen.

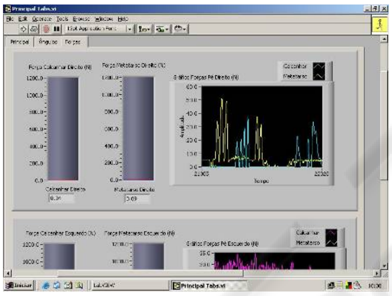

Figure 3: Graphical interface for force measurements.

On the main screen, there is a calibration function. This function should be used when the patient already has the sandals on, but has not stood up yet. The function performs 200 force acquisitions for each load cell, and then calculates the average of these forces. The result is then used as linear coefficients for the calibration curves of the force transducers. This way, the value shown with the patient sitting is zero, despite the applied force for tightening the sandals.

3 RESULTS

The final prototype was mounted on a healthy individual and tested, to check its functionality. The individual's characteristics are: male, 25 years old, height of 1.81m, weight of 69.0kg. Figure 4 shows the subject wearing the instrumented sandals and the shape sensors.

Figure 4: (A) System mounted on healthy individual; (B) Detail of sensor positioning in the sandal; (C) Instrumented sandal.

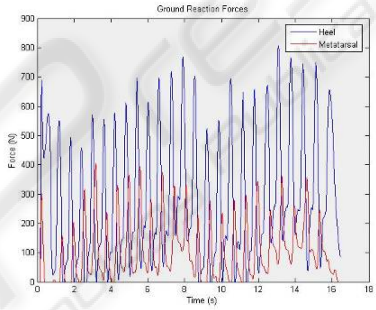

Figure 5: Ground reaction forces on right foot for healthy individual.

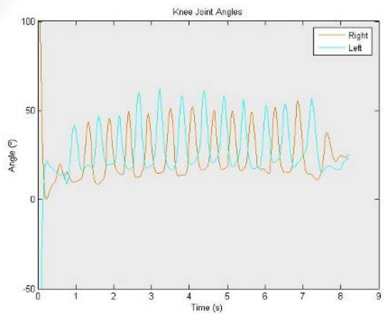

Figure 6: Knee joint angles for healthy individual.

The individual was asked to walk normally in a straight line for about 15m while the values of ground reaction forces and knee joint angles were recorded. Figures 5 and 6 show the resulting values for the ground reaction forces on the left foot, and the knee joint angles, respectively.

For the angle values, the convention adopted was zero for straight vertical position, positive angles for knee flexion and negative angles for knee extension.

4 DISCUSSION

The ground reaction forces observed in Figure 5 show peaks of about 780N on the heel, which correspond to 113% of the individual's weight. On the metatarsal area, the peak force values are around 380N (55% of the individual's weight). This may be due to the softness of the sandal sole, which may still absorb part of the applied forces. It's possible to distinguish the gait phases of initial contact (heel force peak), mid stance (heel and metatarsal force intersection), and terminal stance (metatarsal force peak).

The knee joint angles observed in Figure 5 follow the pattern presented on the CGA Normative Gait Database (Kirtley, 2006). The waveforms present a repetitive pattern, confirming that the shape sensors did not move during the acquisitions.

5 CONCLUSIONS

The values of ground reaction forces and knee joint angles may be observed during the gait, and the recorded values may be used for further analysis, comparing different styles of gait, or different rehabilitation stages for the same individual.

This system may be used as an alternative to the force platforms. The disadvantages are that it requires some time for donning, and it can only measure vertical forces. But it presents some advantages, such as: the subject may walk freely (within the limitation of the cables), and does not have to step exactly on the load cell, resulting in a more natural gait; also, the system allows monitoring two critical force points for each foot, and not just the resulting force.

Considering the aforementioned advantages, an important possible application for this alternative system is suspended FES-assisted gait. In this case, therapists may follow the recovery of patients undergoing this kind of treatment by analyzing the gait on different stages of the rehabilitation process.
In the future, the recordings of gait sessions may be used as inputs for a closed-loop FES control. The system already has two digital inputs and two outputs, which may be used to trigger an electrical stimulator. Since the software is open, an Artificial Neural Network (ANN) may be programmed to control the FES during gait, using the patient's own recorded data for training. With this implementation, the patient will not need to trigger the stimulation manually, and may direct all the attention to the walking activity.

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