

# Portable system for force measurement of long-term monitoring during human gait

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**Abstract.** The characterization of human gait using force platforms is one of the finest areas of biomechanics. Typically these force platforms are large, fixed and restricted to a laboratory environment. Therefore, a system for monitoring the human gait that is flexible, small-sized and can be installed in the actual sole of a shoe is desirable for many applications. We developed a portable, low-power consumption system to continuously monitor the forces involved. Experimental data were compared with simulated data using finite elements method (FEM). The results showed that the system was stable and adequate to application proposed. Results showed a linearity error of 1.25% on the vertical axis, revealing the smooth functioning of the system. Found deformations in the load cells were compared with the simulated model, showing a difference of 5.8% at most. This system might be helpful to monitor force measurement for the human gait.

## 1 Introduction

Simple everyday activities use correct balance and body orientation to maintain correct posture. When walking, running, lying down, and even stop standing, the muscle system is at all times correcting efforts to maintain balance. Even when an individual is standing still, oscillations occur so that the muscle forces control the human body gravity center. Arrangement of relative angles of all joints involved in the equilibrium position can be called posture [1]. The human gait is known to be one of the most complex forms of all human *activities* and its study has a multidisciplinary nature, involving many areas [2], [3], [4], [5], [6], [7]. Research on human gait may help determine some parameters that can be used as markers of health [8]. Disorders in human gait can affect different age groups and are related to various pathologies [9]. The human gait is a complex phenomenon not usually described analytically, because there are a large number of variables to be studied. Thus, the human gait analysis through the soil reaction, stepped reaction, among others simplify the study process and makes possible data acquisition for its characterization.

The development of force platforms is important in human gait analysis. Force platforms can be used in the study of human motion, to assess tremors related to specific psychopathological disorders, as well as in the study and development of new treatments [10-15]. The measurement of the foot-to-floor force interaction is admitted as a great value for locomotion analysis, both in itself and in connection with the measurement of kinematic data of body segments. Foot-to-floor interaction is commonly studied using force and pressure measurement systems. Force platforms are widely used both in posture and gait analysis [16-19]. Typically, the force platforms are medium to large rectangular plates fixed to the floor inside the laboratory or clinic. Faced with the range of sensors and transducers for researches and for automotive and industrial applications, the load cells are considered one of the most important transducers in the market. Basically, a load cell measures deformations caused by different kinds of forces. The main reasons for using load cells are related to its high response fidelity, low cost, small size (even for heavy loads' applications), and to allow dynamic measurements. Typically, in the load cells, in which the elastic element undergoes small deformations, the extensometer is the sensor normally used.

The object of the study was to develop a portable force-monitoring system to continuously measure the forces during the human gait. The load cell can be fixed into a shoe sole, for obtaining more flexibility, which is, allowing testing in environments other than the laboratory or clinic.

## 2 Materials and Methods

### 2.1 System Design

Figure 1 shows the blocks diagram of the force measurement system. The system consists of three parts: a transducer, a signal conditioner and a computer. The first part (transducer) was developed to support an adult human (until 180 kgF), standing or moving. To indicate the applied forces, eight load cells for measurement were built. Figure 2 shows the geometry of the developed load cell (mass: 72.7g and volume: 9.27 cm<sup>3</sup>). Each platform is formed by four load cells (steel: AISI 420C) – see Figure 3. The sensor was a strain-gage PA-09-062AG-350L, self-temperature compensation for steel, 2.13 gage factor, and 350Ω resistance (Excel Sensors).

In the remaining parts of the system, was developed a signal conditioner and utilized a computer for acquisition (ADC – National Instruments USB6008, sampling rate of 256Hz) and processing of signals (Matlab and LabVIEW).

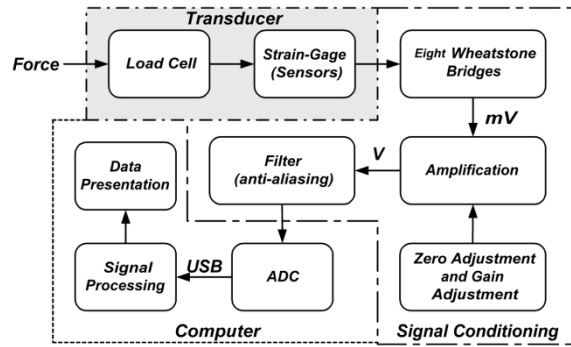


Fig. 1. Block diagram of the experimental system.



Fig. 2. Load cell: photo.

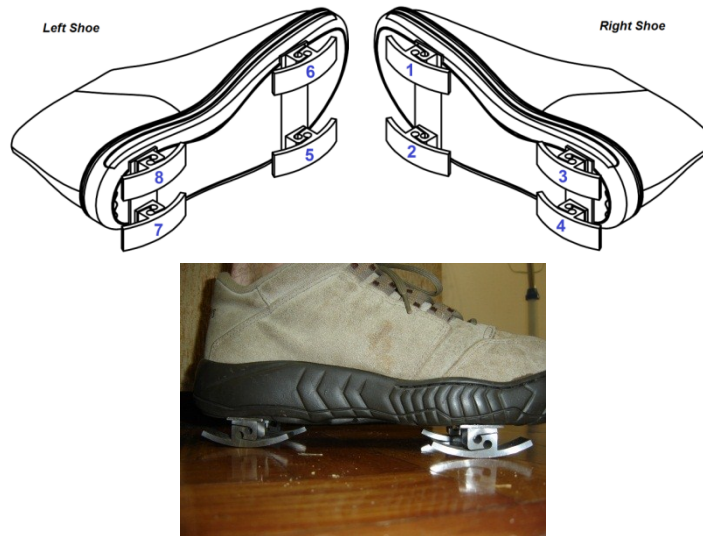


Fig. 3. Platform: (a) sketch and (b) photo.

## 2.2 Circuit Design

Figure 4 shows the circuit design for measuring the force during the human gait. The generated force was conditioned by the instrumentation amplifier (AD620, low drift, low power instrumentation amplification, Analog Devices), voltage reference (REF02:  $+5V \pm 0.2\%$ , Texas Instruments) and filter circuit. The resistance of the transducer was  $350 \Omega$ . The bridge circuit (two active sensors and two passive sensors (dummy sensor)) provides the advantages of temperature compensation and higher sensitivity. The gain of the instrumentation amplifier was programmed to be 894 ( $R_G = 56\Omega$ ).

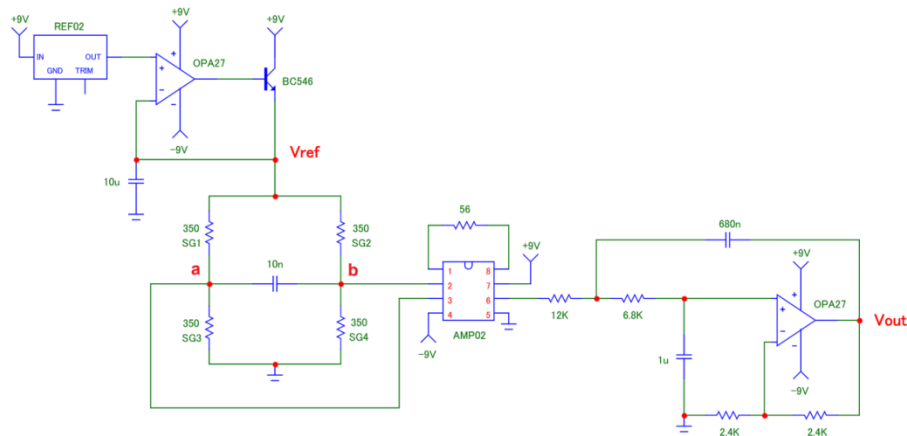


Fig. 4. Circuit design for measuring the force during the human gait.

The low pass filter (Butterworth) with a cut-off frequency of 20 Hz was designed to make the force measurement more clear without contamination. The gain in this stage was designed to be 2 (6.02dB). The current to each strain-gage is approximately (1):

$$I = \frac{V_{ref}}{2R_{SG}} = \frac{5}{2 \times 350} = 7.14 \text{ mA} . \quad (1)$$

and  $V_{out}$  given by (2):

$$V_{out} = 2 \times 894 \times V_{ab} = 1788 V_{ab} . \quad (2)$$

The output voltage of the Wheatstone bridge ( $V_{ab}$ ) and reference voltage  $V_{ref}$  (3):

$$V_{ab} = \frac{K}{4} (-\varepsilon_1 \gamma + \varepsilon_2 - \varepsilon_3 \gamma + \varepsilon_4) V_{ref} . \quad (3)$$

As the deformations ( $\varepsilon$ ) of passive sensors is negligible, we have, approximately (4) and (5):

$$V_{ab} \cong \frac{K}{4}(\varepsilon_2 + \varepsilon_4)V_{ref} . \quad (4)$$

$$V_{out} \cong 1788 \frac{K}{4}(\varepsilon_2 + \varepsilon_4)V_{ref} . \quad (5)$$

The circuit (Figure 4) was battery powered (9V), with low-power consumption. Figure 5 shows the chain of the proposed measure.

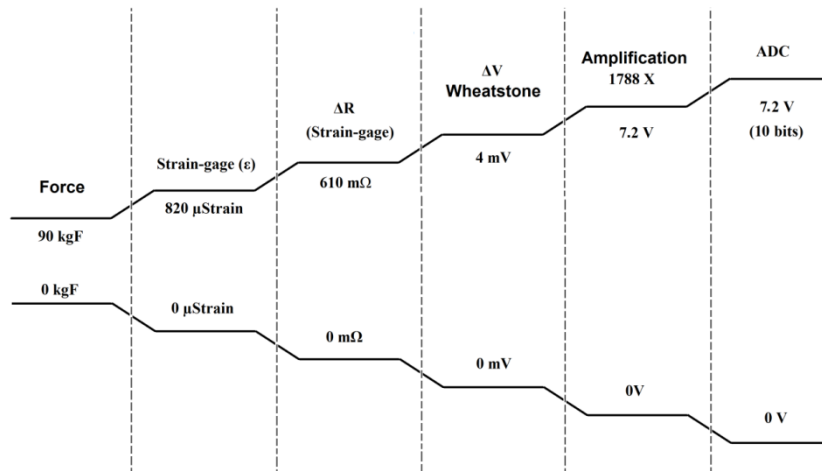


Fig. 5. Chain of the proposed measure.

Dynamic margin (DR) is given by Equation (6):

$$DR = \left| \frac{7,2V-0V}{0,01V} \right| = 720 . \quad (6)$$

Therefore, the input resolution ( $ir$ ) is (7)

$$ir = \left| \frac{90kg-0kg}{720} \right| = 0,125 kg . \quad (7)$$

soon, with a 10-bit ADC (rounding) – Equations (8) – (9):

$$2^n = 720 . \quad (8)$$

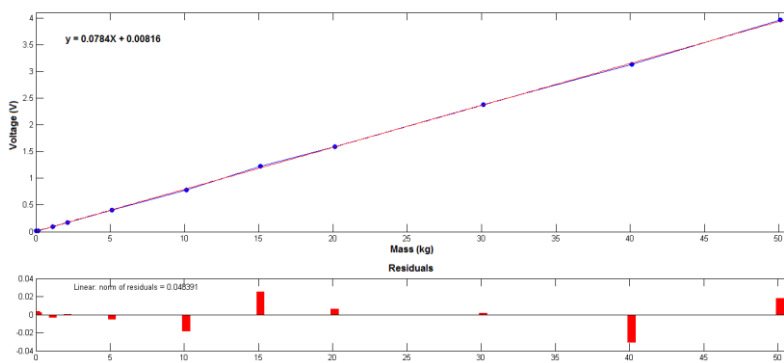
$$n = \frac{\log 720}{\log 2} = 9.49 . \quad (9)$$

### 3 Results and Discussion

#### 3.1 System Performance

The device present low power consumption. Four batteries (9V) were able to make the device work continuously for about 1 day and were sufficient for the purpose of long-term portable usage. The cut-off frequency of the low pass filter was 23 Hz, which almost corresponded with the ideal value of 20 Hz. But this fact is not a problem, because the tests are carried around 5Hz.

Figure 6 shows, for example, the relationship between voltage and mass (transfer function of a load cell – static calibration). Table 1 shows experimental data from all the load cells.



**Fig. 6.** Experimental transfer function: load cell #6.

**Table 1.** Experimental data from all the load cells.

<i>Load Cell</i>	<i>Experimental Transfer Function</i>	<i>Sensi- bility</i> ( $V/kgF$ )	<i>Reso- lution</i> ( $kgF$ )	<i>Linearity error</i> (%)
1	$y = 0.080x - 0.001$	0.080	0.125	1.25
2	$y = 0.039x + 0.003$	0.039	0.253	0.25
3	$y = 0.078x + 0.007$	0.078	0.128	1.25
4	$y = 0.068x - 0.015$	0.068	0.146	0.86
5	$y = 0.077x + 0.005$	0.077	0.130	0.25
6	$y = 0.078x + 0.008$	0.078	0.128	0.75
7	$y = 0.077x - 0.007$	0.077	0.129	1.25
8	$y = 0.079x + 0.010$	0.079	0.127	0.50

Measures in Table 1 was used MDM220 (ECB) multimeter (4½ digits and an uncertainty of  $\pm(0,05\%+1\text{LSB})$  in the range of 20 V). Soon, the uncertainty for all measurements is (10):

$$\text{uncertainty} = \sqrt{(0.05)^2 + (0,1)^2} = \pm 0.112 V. \quad (9)$$

### 3.2 Comparison of system and trials

Assays were performed with this force platform and with a commercial force platform (OR6-5, Advanced Mechanical Technologies, Inc). The results showed no significant difference, thus demonstrating that the load cell proposed in this article may be used in trials of human gait. To perform the experiments, a volunteer with approximate mass of 92.5 kg, used during the human gait, the two force platforms. The signal conditioning and ADC were placed in a backpack carried on the back of the volunteer. With the data processing, the system displays the traditional waveforms representing the behavior of the human gait (see Figure 7).

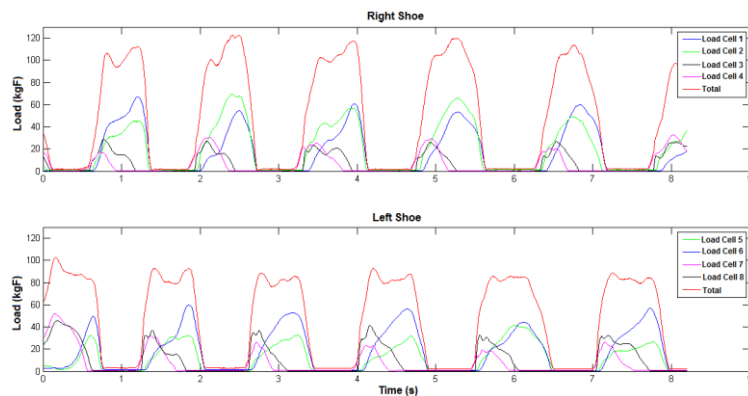


Fig. 7. Example result: a period of human gait.

## 4 Conclusions

In this study, the proposed system was stable enough to monitor the forces during human gait. The Transfer functions of the load cells showed high consistency. The calibration revealed that the actual characteristics were similar to those of an ideal system. The developed system in the study is easy to apply and operate for force monitoring during human gait. The results for the proposed system showed that it is portable and allows flexibility in its use, not being restricted to the laboratory environment. Comparative trials have shown that the responses of the proposed project are appro-

priate to the intended area. The transfer functions for each load cell, exhibited a linearity error of 1.25% maximum for the load cells 1, 3 and 7.

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