# AN INSOLE SHEAR FORCE MONITORING SYSTEM FOR OPTIMUM ALIGNMENT IN LOWER LIMB AMPUTEES

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Lower limb alignment plays a crucial role in providing basic walking functionality in a natural manner as well as the appearance of the gait to amputees. Presently, amputee gait monitoring still remains largely in laboratory scale. There is an urgent need to monitor gait in outdoor activities. Anterior-Posterior (AP) shear force is the most sensitive parameter in prosthetic alignments. A force plate can measure the shear force but it is expensive and can only be available in a gait laboratory. To monitor gait outside the laboratory, a miniature, portable and wearable insole shear force monitoring system is proposed. The system can measure AP shear forces as the body load line sways anteriorly and posteriorly. As weight is a crucial issue, light weight material is used for the sensor housing of the system. Two arrays of force sensitive resistors (FSR) are used to measure the AP shear force as the load line sways. Two insole shear force sensors of the monitoring systems are designated to be bonded strongly underneath a shoe at the heel and the forefoot respectively. This paper presents the development of the monitoring system and initial test results.

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## 1. Introduction

This project involves an investigation into achieving optimum alignment of lower limb prosthesis. A lot of research has been carried out in this area in the last decade. However, the parameters and methodology to identify the optimum alignment is still controversial. Some researchers believe that symmetry [1, 2] is the key in searching for the optimum alignment. They tried to look for the symmetry between the sound leg and the prosthetic leg. Both kinematics and kinetics parameters such as angular displacement and rate, temporal, insole centre of pressure (COP), ground-reaction-force (GRF), weight line and load line, joint moments, were vigorously studied. Some believe in stability and minimum energy expenditure [3-5]. Recently some [6, 7] believe in matching roll over shape (ROS) as close as possible to an ideal ROS shape of the foot is the key to a priori alignment. Somehow none of the researchers have claimed confidently that they have found the key of the optimum alignments. Above all, Zahedi [8] proved that the amputees are highly capable to adapt themselves to a broad range of optimum alignments in level walking. And, he had set a coordinate reference system in order to align the prosthesis for both transtibial and transfemoral prosthetic devices. Later, Sin [9] re-examined the accepted range and proposed a non-level walking test which can constraint the range into a much smaller.

Misalignment of the lower limb prosthesis would cause physical damages to the muscular system and musculoskeletal system. Undesired pressure distribution in the stump/socket interface would result in great discomfort, and continuous mechanical abrasion will eventually cause tissue breakdown, bruise, irritation, stump pain and skin problems. Stump skin damages are serious and should be avoided. Furthermore, heavy and consistent dependency on the sound limb would cause undesired pressure distribution to the rest of musculoskeletal system and hence increase in the prevalence of degenerative changes in the lumbar spines and knee.

## 2. Aim

The aim of this work is to develop a monitoring system that measures AP shear force changes upon alignments in the prosthetic ankle tilt/shift. Such a system would correlate the relationship amongst AP shear force and prosthetic ankle tilt/shift to provide an objective means for alignment process to help prosthetists.

## 3. Methodology

Figure 1a and 1b show the working principle of an insole shear force sensor and the decomposition of the body load line in the directions of progression and upward reaction. Two sensors are installed beneath the shoe at the forefoot and the heel (Figure 1c). By the principle of force, a body load line, $W$, is decomposed into the shear force (Eq. (1)) and the upward reaction force (Eq. (2)).

$$\text{Shear Force, } F_H = W \cdot Sin(\theta)$$

Equation. (1)

$$\text{Upward Reaction Force, } F_V = W \cdot Cos(\theta)$$

Equation. (2)

We are interested in the shear force. The sensor housing consists of a sliding block and a pair of static blocks as seen in Figure 1a. Two rollers minimize the friction between the sliding block and the static block. Its dimension is 68 (L) x 30 (W) x 16 (H) (mm). Two arrays of Force Sensitive Resistors (FSR) are attached at both ends of the sliding block.

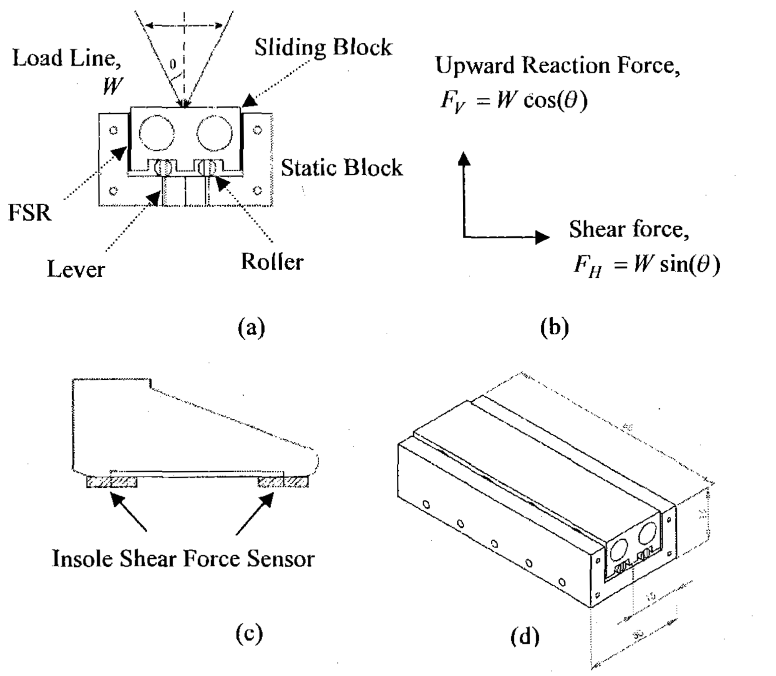



Figure 1: (a) Working principles of Insole Shear Force Sensor, (b) Decomposition of the body load into the directions of progression and upward reaction, (c) Sensors attached to the shoe, (d) 3D design of the sensor housing

Human walking is a cyclic motion and the body load line would sways at approximately $\pm 15^0$ at normal walking speed. The insole shear force is expected to range around 25% of the body load. All data are collected using USB data acquisition card (Part No: PMD-1208LS) and Labview version 8.5.

## 4. Static Characterization of FSR

An inverting op-amp circuitry is constructed. Amplification gain is carefully selected to avoid output saturation at maximum input load. A regulated 5VDC voltage to FSR is needed because the output of a FSR is ratiometric. Figure 2 shows the FSR (FlexiForce, A201-25) and a custom-made test rig for the purpose of static characterization. The puck of the platform is visually adjusted to fit on top of FSR's sensing area. Known dead-weights (0.51Kg each) are stacked onto the platform per reading. Eight FSRs were tested and for eight trials each. Before test, each FSR is pre-conditioned as recommended in the user manual [10].

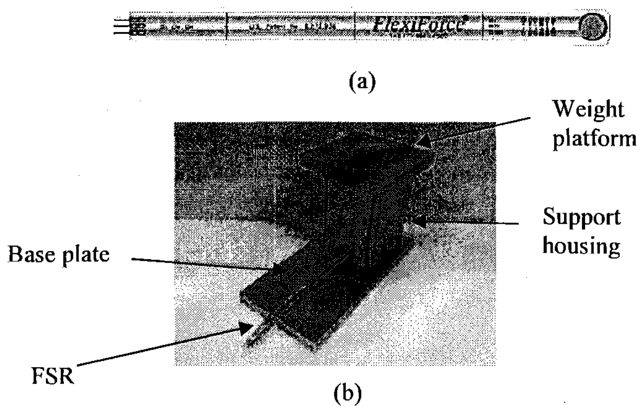



Figure 2: (a) FSR (b) Test Rig

Table 1: Static characteristics and comparison

| | Repeatability error, $\varepsilon_R \pm 2SE$ | Hysteresis Error, $\varepsilon_H \pm 2SE$ | Non-linearity error, $\varepsilon_L \pm 2SE$ | Max. input range (Kg) | Max. output range (V) |
|---|---|---|---|---|---|
| Exp. | $2.40 \pm 0.37\%$ | $3.20 \pm 0.45\%$ | $8.81 \pm 1.60\%$ | 11.29 | $6.79 \pm 0.76$ |
| Spec. | $\leq \pm 2.5\%$ | $< 4.5\%$ | $< \pm 5\%$ | - | - |

The static characteristics are listed in Table 1 and compared with specifications provided by the manufacturer, Tekscan, Inc. Experiment data are described in mean (μ) and standard error of means (SE). Repeatability error and hysteresis error are well within the specification but non-linearity error is obviously larger than given specification.

Mean and two SE( 95% CI) of eight trials are plotted (shown in Figure 3) and by linear regression method, Eq. (3) is formulated, which displays highly fitted coefficient of determination, $R^2 = 0.9929$. SE in regression is 0.1724, which describes the standard deviation of the mean of the predicted output for any given input. Statistically, for any given input, the predicted output is described in Eq. (4).

$$y = -0.5914x - 0.5124$$

Equation 3

$$y\_predicted = y \pm 2 * SE$$

Equation 4

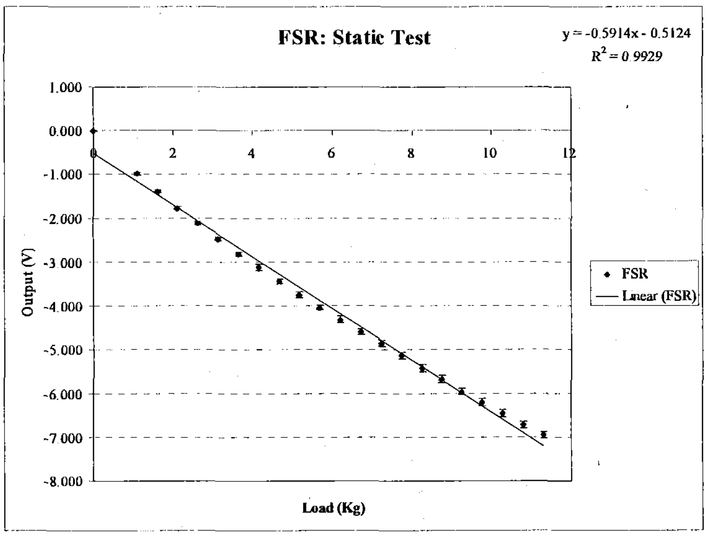
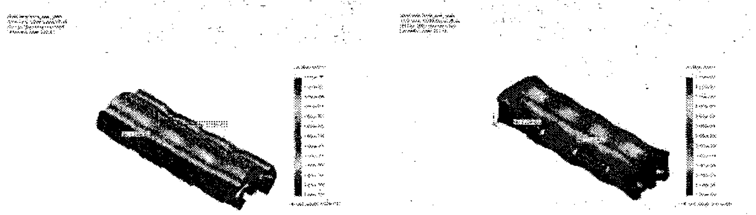



Figure 3: Mean, Standard Error and Linear Regression

## 5. Static Stress Test

Before fabrication, nine iterations of design alterations were attempted to achieve two objectives: the lowest factor of safety (FOS), FOS ≥ 3.0 and total weight ≤ 100g. Weight is reduced by hollowing or thinning each single part while maintaining FOS. The designated vertical load is $W_V = 1000N$ and horizontal load is $W_H = 200N$. Static stress simulation (SolidWorks – COSMOSXpress) as in Figure 4, checks all the designs per iteration and compares the results between aluminium and standard steel. Table 2 lists the simulated results after nine iterations and aluminium is chosen for its light weight.



(a) (b)

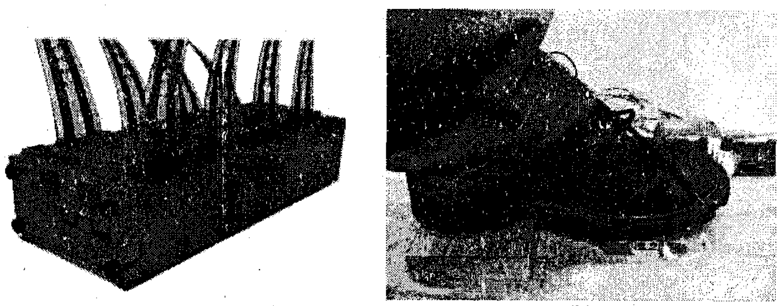
Figure 4: Stress simulation, (a) under vertical load, (b) under horizontal load

Table 2: Static stress simulation result after nine iterations

| Part | FOS ($W_V$) | | FOS ($W_H$) | | Weight (grams) | |
|---|---|---|---|---|---|---|
| | Alu. | S. Steel | Alu. | S. Steel | Alu. | S. Steel |
| Sliding Block | 4.69479 | 51.6522 | 16.5497 | 182.656 | 22.89 | 65.27 |
| Static Block | - | - | 8.39176 | 92.8373 | 23.13 x 2 | 65.27 x 2 |
| Roller | 6.19943 | 69.403 | - | - | 1.22 x 2 | 3.47 x 2 |
| | | | | Total | 71.59 | 204.16 |

## 6. Prototype

A prototype is assembled as shown in Figure 5a. A feasibility study (Fig. 5b) was performed to check if the sensor output would spring back to its neutral reading after unloading, and the direction of sensor output upon forwarding and reversing load. The results were positive. Figure 6 shows the results of the feasibility study on Labview.



(a) (b)

Figure 5: (a) Prototype of the insole shear force sensor, (b) Feasible study

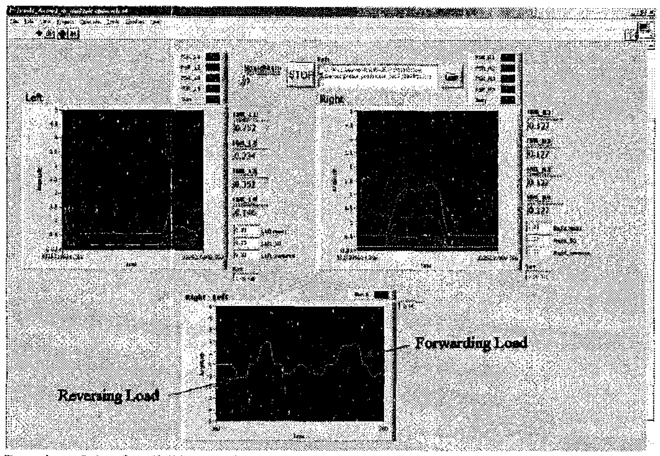



Figure 6: Results of the feasibility study

## 7. Conclusion

To date, a novel monitoring system is partially developed and initial feasibility studies confirmed its functionality to collect the insole shear force in the direction of progression. This system would help in the quantitative assessment of the prosthetic alignment to assist prosthetists in objectively aligning prosthetic devices. The future works include calibration of the sensor over full scale load of 100Kg and $\pm 15^{\circ}$ swags; the shoe mounting mechanism; a light-weight data logger; and signal conditioning and processing.

**Reference**


1. R. E. Hannah, J. B. Morrison, and A. E. Chapman, "Prostheses alignment: effect on gait of persons with below-knee amputations," *Arch Phys Med Rehabil,* vol. 65, pp. 159-162, 1984.
2. E. Isakov, J. Mizrahi, Z. Susak, I. Ona, and N. Hakim, "Influence of prosthesis alignment on the standing balance of below-knee amputees," *Clinical Biomechanics,* vol. 9, pp. 258-262, 1994.
3. J. W. Breakey, "Theory of Integrated Balance: The Lower Limb Amputee," *Journal of Prosthetics & Orthotics,* vol. 10, pp. 42-44, 1998.
4. S. Blumentritt, "A new biomechanical method for determination of static prosthetic alignment," *The Journal of the International Society for Prosthetics and Orthotics,* vol. 21, pp. 107-113, 1997.
5. S. Blumentritt, T. Schmalz, R. Jarasch, and M. Schneider, "Effects of sagittal plane prosthetic alignment on standing trans-tibial amputee knee loads," *The Journal of the International Society for Prosthetics and Orthotics,* vol. 23, pp. 231-238, 1999.
6. A. H. Hansen, D. S. Childress, and E. H. Knox, "Prosthetic foot roll-over shapes with implications for alignment of trans-tibial prostheses," *Prosthetics and Orthotics International,* vol. 24, pp. 205-215, 2000.
7. A. H. Hansen, M. R. Meier, M. Sam, D. S. Childress, and M. L. Edwards, "Alignment of trans-tibial prostheses based on roll-over shape principles," *Prosthetics and Orthotics International,* vol. 27, pp. 89-99, 2003.
8. M. S. Zahedi, W. D. Spence, S. E. Solomonidis, and J. P. Paul, "Alignment of lower-limb prostheses," *J Rehabil Res Dev.,* vol. 23, pp. 2-19, 1986.
9. S. W. Sin, D. H. Chow, and J. C. Cheng, "Significance of non-level walking on transtibial prosthesis fitting with particular reference to the effects of anterior-posterior alignment," *J Rehabil Res Dev,* vol. 38, pp. 1-6, 2001.
10. Flexiforce Sensor User Manual, Tekscan Inc. Dated: 23rd September 2005. RevF