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A Low Cost Instrumentation Based Sensor Array for Ankle Rehabilitation

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1. Introduction

Tactile sensing technology has made significant progress towards the development of devices where robot fingers must have the ability of multi-dimensional tactile sensing in order to perform grasping and manipulating tasks (Chi & Shida, 2004); (Webster, 1998); (Nicholls & Lee, 1989); (Tarchanidis & Lygouras, 2001); (Da Silva et al., 2000). Therefore, many researchers have tried to develop various types of tactile sensors by applying MEMS technologies which usually rely on the measurement of pressure or force on a sensing element (Hasegawa et al, 2007). A variety of different types of sensors have been used, including resistive strain gauges, piezoelectric film, infrared displacement sensors, capacitive sensors, sensors detecting conductance, magnetic, magnetoelectric and ultrasonic sensors. Frequently large numbers of sensing elements are built into an array and the outputs of those sensors are processed, often in conjunction with a mathematical model, to give an assessment of the contacting object. Nowadays, force sensing becomes an important component for diver applications in biomedical applications and orthopedic rehabilitation. Thus, tactile sensors have been used in hand clinical evaluations and foot rehabilitation (Da Silva et al., 2000); (Mascaro & Asada, 2001); (Boukhenous & Attari, 2007); (Attari & Boukhenous, 2008). Human tactile sensing is achieved by means of at least four different types of receptor cells (Jayawant, 1989); (Cowie et al., 2007) and is used to feel, grasp and manipulate objects, and to assess attributes such as shape, size, texture, temperature, hardness, discontinuities such as holes or edges, and movement, including vibration. Reston and Kolesar (Reston et al., 1990) described a robotic tactile sensor manufactured from piezoelectric polyvinylidene fluoride (PVDF) film. It was not the best choice for finger mounted tactile sensors due to its limited load range and inability to measure static forces. Beebe and al (Beebe et al., 1989) developed a force sensor based on a silicon diaphragm structure instrumented with piezoresistors in Wheatstone bridge configuration. The applied force is distributed across the diaphragm via a grasping solid dome and mounted on rigid substrate with an excellent performance characteristic. In this paper a low cost tactile sensors array for the measurement of hand grasping forces is described in a first step. A second step is dedicated to the study of two-dimensional reaction forces distribution of foot during rehabilitation in the case of ankle sprain. The sensor element of the array is an easy structure based on the use of low cost Hall device and a general purpose polymer (polysiloxanes). First the elastic polymer is studied to show its ability in building such

sensing element. After mounting the sensors, the outputs signals are conditioned, filtered and then digitized with a high resolution data acquisition card. A static calibration test has been fulfilled to estimate the degree of linearity. Preliminary measurement has been carried out concerning the fingertip forces grasping of hand during holding objects and the distribution of impacts forces during foot contact.

2. Principe and sensor design

For the design of the sensor element, a Hall Effect sensor UGN3503 from Allegro Micro-Systems was selected. This sensor is used for measuring magnetic flux densities with extreme sensitivity and operates well in the temperature range from –20°C to +85°C and in the frequency range from DC to 23 kHz. This device is widely used for measuring linear position, angular position, velocity and rotational speed. Hall sensors are also commonly incorporated into CD-ROM drive, hard disk drives, automotive ignition, electrical current sensing and ABS braking systems as they are robust, unaffected by dirty environments and low-cost (Ripka & Tipek, 2007). In contrast to other magnetic sensors, the manufacture of Hall magnetic sensors does not require special fabrication techniques as they are compatible with microelectronics technology. Most of the sensors are low-cost discrete devices but an increasing proportion now come in the form of integrated circuits. The integrated Hall magnetic sensors usually incorporate circuits for biasing, offset reduction, temperature compensation, signal amplification and signal level discrimination. The most advanced Hall sensors incorporate digital signal processing and are programmable such as HAL800 from Micronas (Bushbaum & Plassmeier, 2007). The considered sensor element is constructed by placing a magnet which produces a constant magnetic field nearby the selected Hall sensor. The layer of thickness d between the magnet and the Hall sensor is realized with an elastic polymer materiel (Fig.1). Special care was dedicated to the positioning of magnet in the direction of the surface area of sensing in order to reduce the nonlinearity of the tactile sensor (Ehrlich, 2000); (Kyberd & Chappel, 1993). After the placement of the different layers composing the whole sensor element, a thin twisted pair wire is soldered to the Hall sensor as the voltage produced is at low level and need low noise amplification.
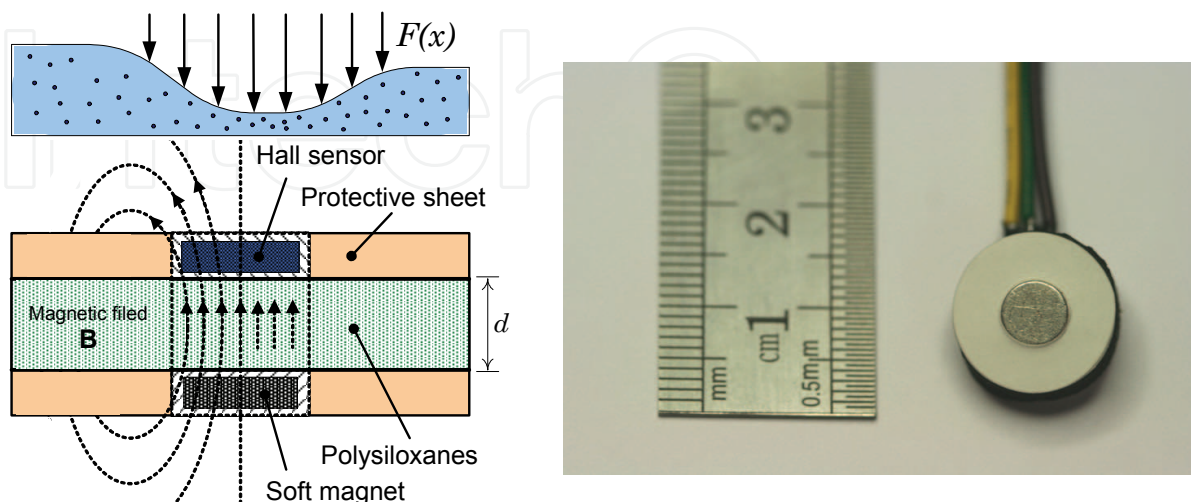

Fig. 1. The sensor element principle and realization

First, the elastic polymer (polysiloxanes) and a piece taken from mouse mat were studied to show the possibility of using this material in building the sensing element. A test bed with Lutron FG-5000A was performed for this purpose and experimental data are reported in Fig.2 for the two chosen materials.
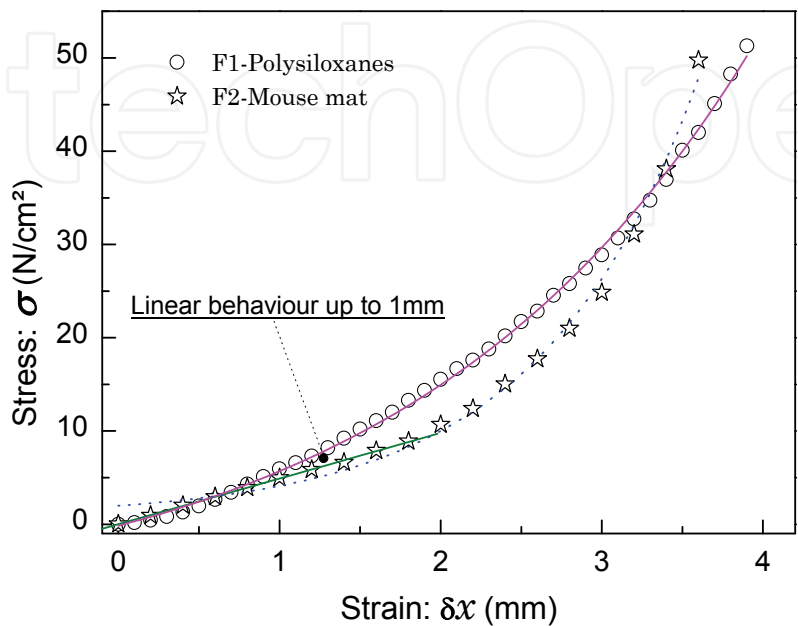

Fig. 2. Characteristics of the materials

For the second material (mouse mat) a strong nonlinear behavior was observed for strain greater than 2 mm. For strain up to 2 mm, the characteristic was quasi linear. The second material exhibits a better monotony with soft nonlinearity. As a calibration curve the following exponential growth was found with a correlation coefficient of about 0.997:

$$F = \beta + \alpha \times \exp(\delta x / k) \tag{1}$$

A more precise calibration curve was obtained with a third-order polynomial with a correlation coefficient of about 0.999, thus:

$$y = F(\delta x) = a_0 + a_1.\delta x + a_2.(\delta x)^2 + a_3.(\delta x)^3 \tag{2}$$

As a nonlinear property is found for the studied material, a software routine was implemented after digital signal acquisition to perform nonlinearity correction. From the calibration curve of the sensor an equi-spaced 1-D look-up table is created and a quadratic interpolation was used (Attari, 2004); (Dias Pereira et al., 2001) whose curve passes through three points $(y_{k-1}, \ \delta x_{k-1})$, $(y_k, \ \delta x_k)$, $(y_{k+1}, \ \delta x_{k+1})$,

$$\delta x = \delta x_k + (y - y_k)f[y_{k-1}, y_k] + \\ + (y \text{-} y_{k-1})(y - y_k)f[y_{k-1}, y_k, y_{k+1}] \tag{3}$$

with the second divided difference given by,

$$f\left[y_k, y_{k+1}\right] = \frac{\delta x_{k+1} - \delta x_k}{y_{k+1} - y_k}$$

$$f\left[y_{k-1}, y_k, y_{k+1}\right] = \frac{f\left[y_k, y_{k+1}\right] - f\left[y_{k-1}, y_k\right]}{y_{k+1} - y_{k-1}} \qquad (4)$$

3. Signals conditioning and experimental

The outputs signals issued from the sensors elements are carried onto a low level instrumentation amplifier (AD622, *Analog Devices*) with low offset voltage, low noise and high CMRR. After analog conditioning, these signals are filtered with a second order Butterwoth active filter and sampled and digitalized with a commercial National Instrument data acquisition card (*DaqBoard 1005*) and then fed a PCI PC bus. Fig.3. show the analog and digital part of the prototype circuit which is directly connected to each sensor element where the output signals are multiplexed with the circuit included in the data acquisition card. First step is to perform the static calibration characteristics by applying a variable force from 1 to 10N provided by a test bed (*Lutron FG-5000A*). Fig.4 shows outputs signals from five sensors elements. Least squares linear regression were performed to compute the estimated linear calibrating curves and to determine the sensor sensitivity for each sensor. After analyzing the sensors data, a linearity was observed for the range [0-10N] with a correlation coefficient greater than 0.99. For forces up to 10N the responses become less accurate against linearity shape and correction based on the method described above (Sec.2) was performed for further investigation, for instance in 2D stress measurement for foot reaction forces distribution. For dynamic experimentation two tests in real environment have been realized.
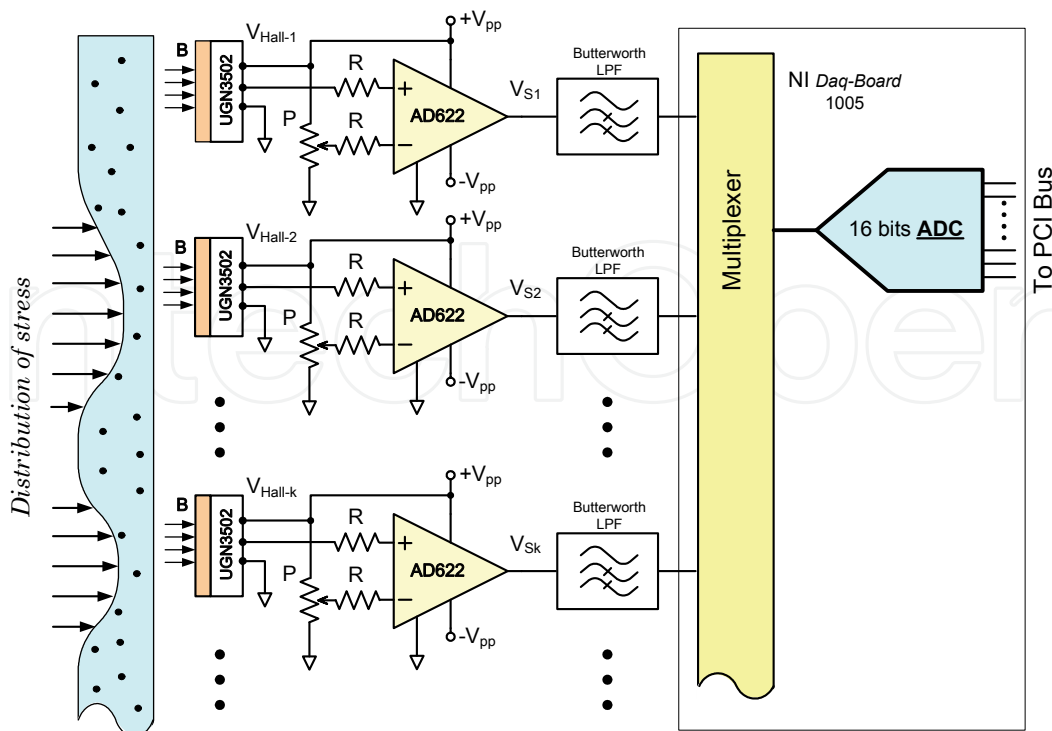

Fig. 3. Conditioning circuit array and data acquisition

3.1 Test during holding objects

For the test five sensors element are bonded onto the fingertips of human hand (Fig. 5). Outputs signals are observed and a software program is developed to analyze the fingertips movement during holding objects. Fig.6 shows the response corresponding to grasping of the thumb, index, middle, ring and little fingertips during holding a bottle of mineral water. The experimental results show that the changes of dynamic fingertips force affects the transducers in the contact phase measurement. The thumb, index and middle are the fingers that give the highest signal level as they exert high pressures regarding the two other fingers. This observation is in concordance with the biomechanics of hand which verify the feasibility of the proposed sensors arrays.
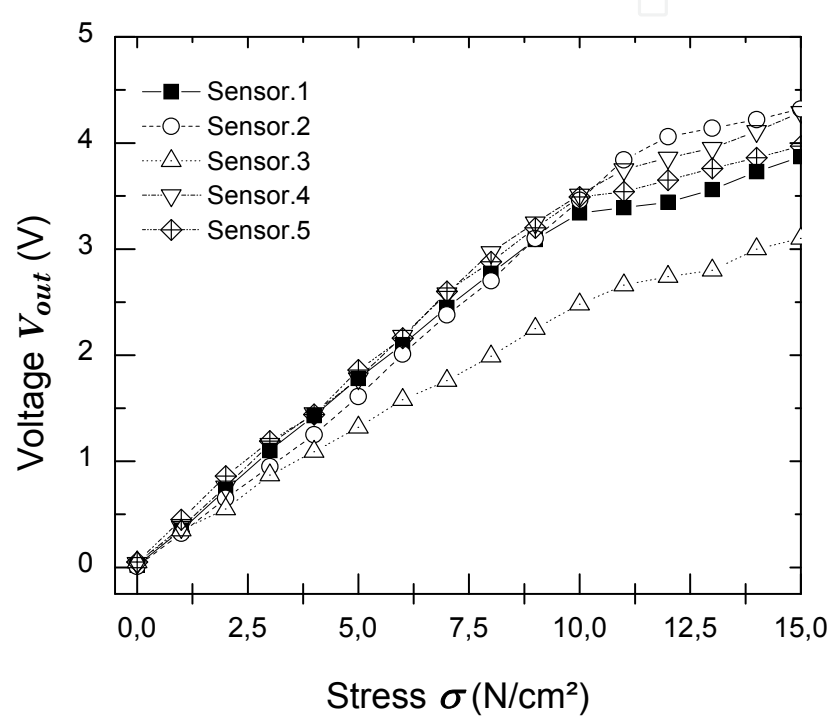
Fig. 4. Static calibration

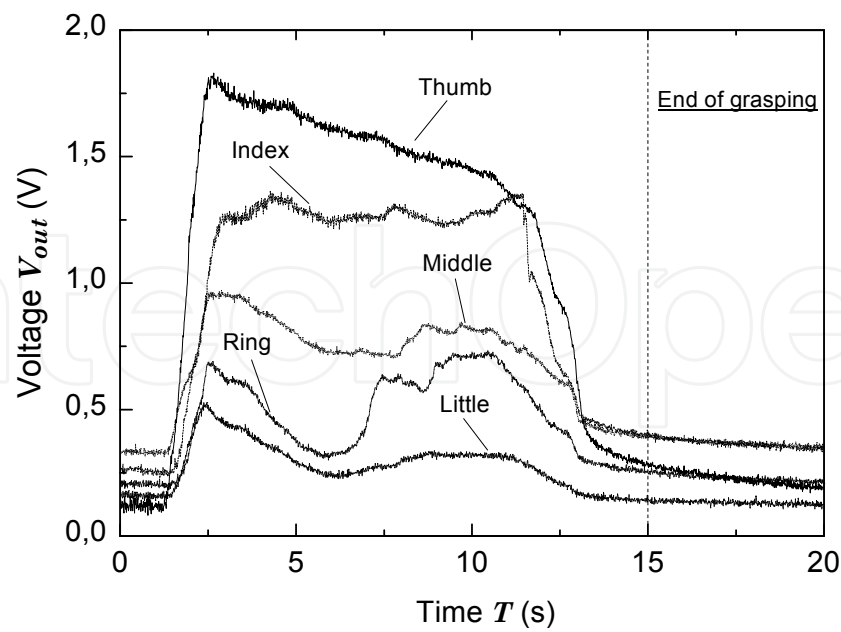
Fig. 5. Tactile sensors bounded on fingers hand

Fig. 6. Outputs signals of transducers during holding

3.2 Test for ankle rehabilitation

Second dynamic measurement in real environment has been carried out with eight realized sensors which are bonded onto a flexible material as foot shape (Fig. 7). Fig. 8 shows the apparatus constructed with wood beech dedicated for the rehabilitation of ankle. Fig.9 shows the response corresponding to eight tactile sensors distributed on the insole surface during an experiment of ankle rehabilitation. The experimental results for 30s recording show clearly the frequency swing of the wood substrate. Also, a delay time is observed for example between sensors S1 and S8 during foot swing where the whole body is maintained stable with one foot. This observation is in concordance with the geometry of the placement of sensors and it is obvious to show that the time delay is approximately half time the time of swinging, thus,

$$T_{Delay} \approx \frac{1}{2} T_{Swing} \tag{5}$$
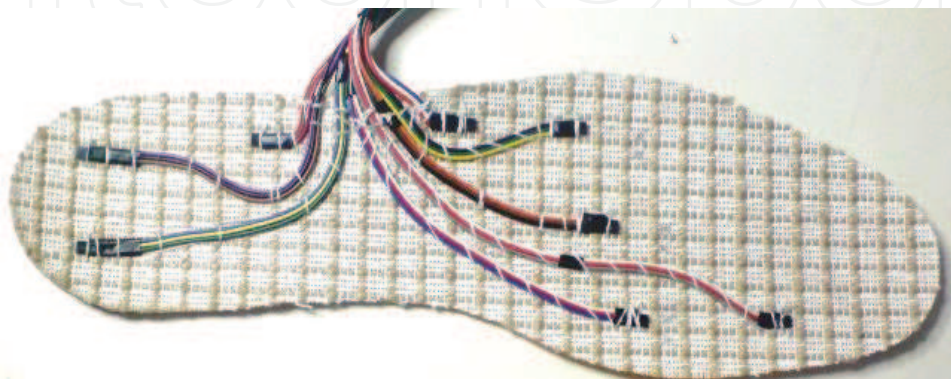

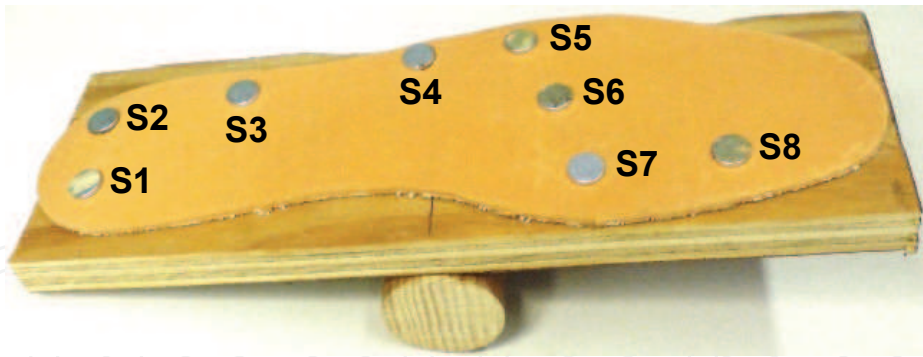Fig. 7. Placement of eight tactile sensors
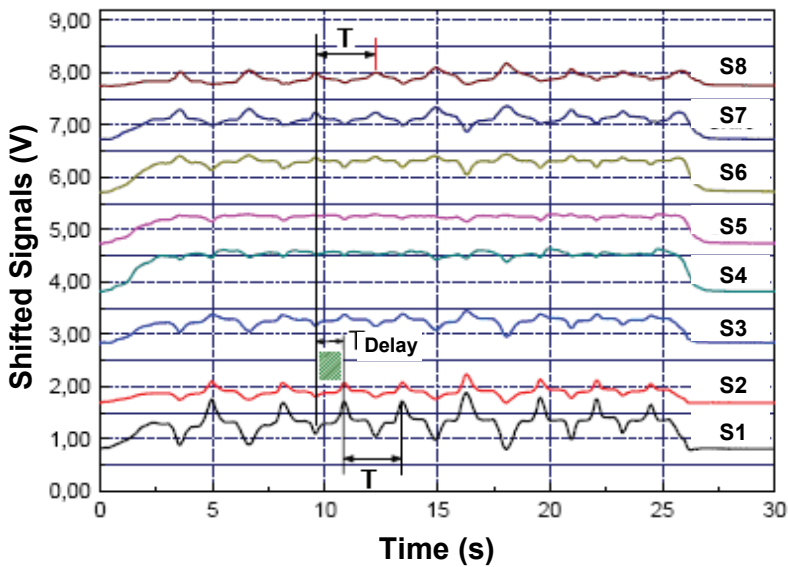
Fig. 8. Apparatus for ankle rehabilitation

Fig. 9. The eight recorded signals

Futures investigations are oriented toward the realization of embedded bioinstrumentation system for the measurement of foot reaction forces for a dedicated balanced platform. This one will be the essential part of the test bed system for the determination of force shape of foot during the rehabilitation of ankle. Fig. 10 shows the principle part of the whole system which consists on positioning a numbers of sensors elements on a special platform fit with dimension of a standard foot. The number of sensors will be determined with resolution required for the foot reaction forces study (Boukhenous et al., 2006). For better flexibility of data acquisition with high special resolution, the HAL800 digital programmable Hall Effect device is preferred. The proposed printed circuit board (PCB) for the realizing of the whole 2D sensing system is shown in Fig. 11. Notice that the number of signals outputs pads is equal to the number of sensors elements. Also, a special care will be considered in positioning precisely the Hall devices with taken into account shielding and grounding of the whole PCB. An epoxy resin will be deposited carefully on the sensors array in order to standardize the first layer against the elastic material.
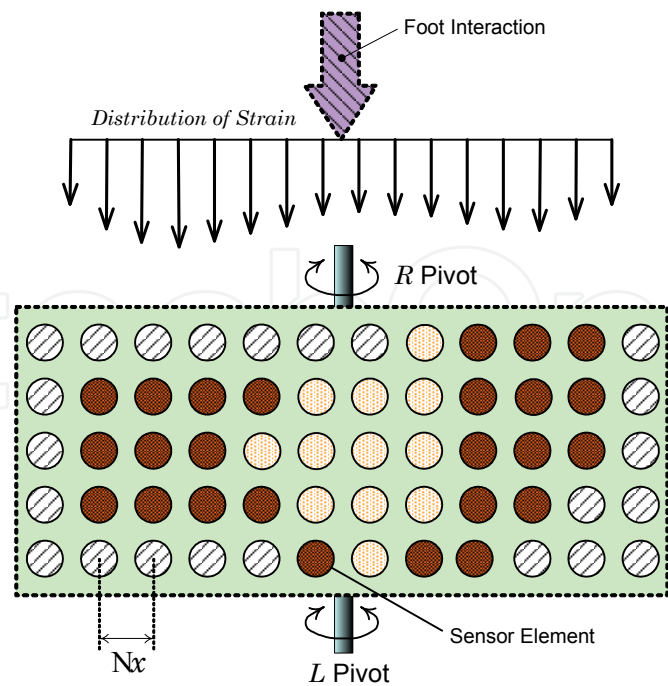
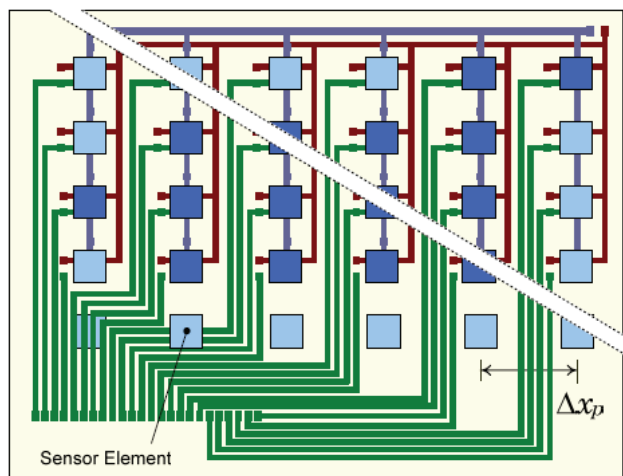Fig. 10. Tactile sensors array for ankle rehabilitation

Fig. 11. Placement of sensors elements in a rigid PCB

4. Conclusion

In this paper a low cost tactile sensors array for biomedical applications are presented. Each sensor element was constructed separately and based on the use of Hall sensor devices. The sensors were calibrate and trimmed before proceeding to the experimental tests. A dedicated analog signal processing was designed and realized according to the specificity of the realized sensor. Accurate settings have been achieved by offset and gain trimming for zero crossing and required sensitivity. Outputs signals from the conditioning circuit of the eight transducers are coupled to a high resolution data acquisition card. The software program developed analyzes and calibrates the multisensors signals. Dynamic experimentation for fingertips grasping of the hand during holding an objects and the

distribution of impacts forces during foot contact for ankle rehabilitation shows a satisfactory response and verify the feasibility of the proposed sensors array. After analyzing the sensors, the data found in the range [0-10N] is the optimized interval for best linearties. Future works are focused toward an intelligent calibration and processing of the acquired signals using dedicated analog processor and FPGA implementation of a matrix of sensors elements for the monitoring of ankle rehabilitation.

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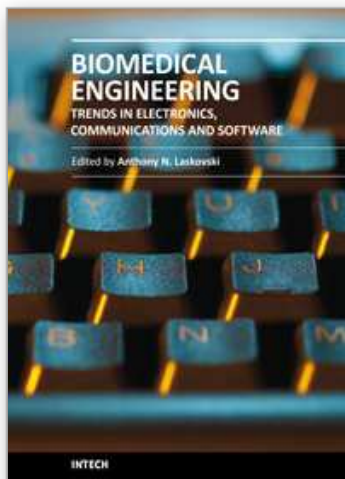
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