Novel Design for Foot Pressure Measurement using Opto-Electronics in Rehabilitation

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DISCLAIMER

This thesis is the result of the author's original research. It has been composed by the author and has not been previously submitted for examination which has led to the award of the degree.

Signed

Date:

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ABSTRACT

The purpose of this research has been to explore the application of a new technology to design a new uniaxial transducer for measuring the plantar pressure of the foot. The measurement approach used here is an optical system that could be incorporated between the foot and the shoe via an in sole. The information obtained can be used to assess the pressure and the kinetic variables that will help the physician to understand the patients' locomotor system better. For this project, a mechanical arrangement has been designed for substantiating the concept of the project. An optoelectronic circuit consisting of an infra-red emitting diode and a photo-detector is placed inside the mechanical device. The emitter diode emits 940 nm infra-red rays and strikes on a reflective surface where the compressive loading changes the refractive angle and the path length of the transmitted light, thereby varying the intensity of light reaching the detector. The change in intensity of light detected will be proportional to the change in voltage and the recorded voltage is observed in a voltmeter or an oscilloscope. Static compressive loading using load plates and an INSTRON is performed here to initially validate the proof of concept. Primary results show that the electronic circuitry has responded to the compressive loading. The rubber used instilled a non-linear response of the transducer, but nevertheless the concept is validated for miniaturisation and incorporated into applications.

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1 INTRODUCTION

1.1 INTRODUCTION

Human gait plays an important role in maintaining the posture and flexibility of the human body. Most of the postural features depend on the pace and the manner in which we walk, thereby determining the flexibility and the distribution of the load to the lower part of the body. If the load distribution of the foot is not properly assessed and corrected, it can cause gait deviations (abnormality in walking pattern) and later instability to the limb system. Injuries too can cause change in the gait kinematics and can alter the gait pattern from the normal style. The primary cause of concern will be damage to the musculoskeletal system and to the joints of the lower limb.

To understand gait deviations, it is necessary that the deviations are analysed using various mechanical systems such as force plate or force transducers. Based on the kinetics of motion, the main parameters assessed are; force, pressure, mass, moment and friction (Billing, Nagarajah, & Hayes, 2002). There has been a range of techniques where gait motion analysis is done which have allowed the physicians to identify the gait problems. Generally the plantar pressure forces are measured using force plates and by measuring the foot and ankle movements. The pressure application and force bearing are mainly focused on the ankle and knee joints, as it is the epicentre for the weight bearing and pressure distribution points. Most of the forces measured using these techniques are the reaction forces or Ground Reaction Force (GRF), which spreads across the plantar surface foot during contact. However, here the main area of concentration will be focused on the vertical forces avoiding the medio-lateral shear forces.

Studies by (Chedevergne & Faivre, 2007) showed that the main pressure points in the human foot are concentrated around the lateral surface of the foot. Except the arch of the foot, the plane surface exerts pressure starting from the heel, transferring towards the external side then the meta-tarsal area and finally with the foot landing on the hallux. This is the how the normal gait pattern of the human foot is. A pictorial representation has been shown in fig.1.2. A study conducted by (Chedevergne and Faivre, 2007) also indicated that eight anatomical positions were well suited to estimate the interaction of the foot to the ground (Fig 1.1).

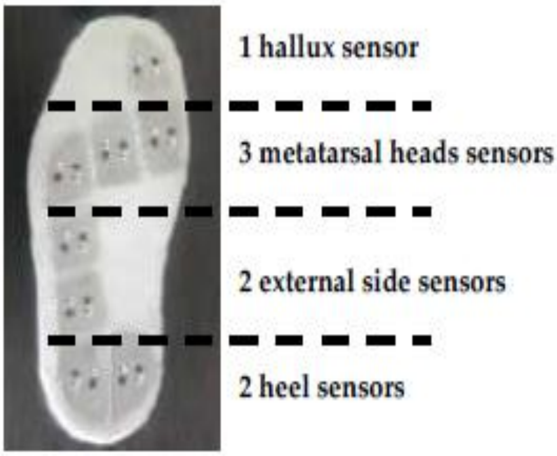

Figure 1.2 Anatomical position for placing sensors (Chedevergne & Faivre, 2007)

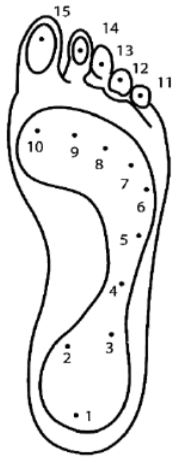

Figure 1.1 Foot plantar pressure map.(Abdul Razak, Zayegh, Begg, & Wahab, 2012)

Preceding the use of pressure measurement devices, all the clinical testing and the evaluation trials were done using the force plate system. Force plates provide information on the intensity of the load applied between the foot and the force while performing a physical task. However, the accuracy of these plates is vital and can have a profound effect on the final evaluations. Hence it is necessary to calibrate the plate time and again. But Cappozzo, in his report has stated that the sustainable techniques for the calibration of the force plates are not easily available for implementing. (Superiore & Giacomozzi, 2010)

1.2 GAIT ANALYSIS

Clinical gait analysis is the investigation of the walking pattern. Many clinicians assess the dynamic pressure distribution and make decisions based on the data. Gait analysis is usually carried out in a well-set laboratory equipped with motion sensors and fully computerised technology to detect and record the gait motion. A monitor displays the data after being processed by unique software designed by the respective companies such as Pedar x/B Basis software (Novel, n.d.-a), Pedar x/E Expert software (Novel, n.d.-b), Footscan software from RSScan (RSScan Lab, 2006). The data obtained may contain information about the intensity of forces applied, application of pressure at certain points, movement of lower limbs et cetera. Analysing the gait motion is approached by two methods. The conventional method being the force plate method, where the subject has to walk across a walkway with sensors placed on the path, detecting the motion. It is repeated 3-5 times, but maybe even more for repeatability. The advantage of using this kind of a system is that large number of sensors can be used thus providing greater resolution. Since all the sensors are placed on the walkway, true vertical placements of the foot were recorded. However, the problems associated with this system is that it needs large number of steps for data collection and 'targeting' by the patient that cause the patient to change their walking pattern to be in contact with the platform (Mueller K, Cornwall M W, 1995). The second method is the insole technique, where an insole embedded with the transducers is placed within the shoe of the subject and made to walk normally. The motion is recorded within less number of steps, which is an advantage over the force-plate system. This method can remove the problem of targeting. Yet, the resolution will not be achieved, as there is limitation to the number sensors placed within the sole. In addition, individual sensors can be damaged due to repetitive loading. In a study conducted by (Kirkeide K, Carmines D, Abel M, 1998) they showed that use of in-shoe method was efficient and easy for data collection.

1.3 IN-SHOE MEASUREMENT CONDITION

The sole of the foot is the interface between the foot and the ground. Pressure variances cause changes in stress patterns and can inflict temporary or permanent changes. At the foot-shoe interface, forces can produce tension, compression, bending, shear and torsion (Billing et al., 2002). Since all this depends on the direction and point of application of the forces and the way the transducer detects these forces, placement of the transducer for pressure measurement is important, as the output readings greatly depend on the positioning of the transducer. Erroneous results can arise if either the transducer is not arranged properly or the foot is not placed properly over the transducer. Therefore, placement of the foot is also another important factor for measuring forces. Moreover, due to distinctive design of the shoes the positioning of the sole can vary. It can be interpreted that the analysis of stress across foot is complex and very dynamic in nature, as the sole undergoes changes and design and selection of the material is vital in developing a transducer.

Approach to introduce an in-shoe between the foot and shoe interface can bring in practical issues also. Since it acts as a barrier between the foot and the measuring surface, most of the characteristics mentioned are lost. Hence, it should be made sure that the thickness of the sole should be limited as possible. Being thick in size can be a concern for the patient. Thus, the in-shoe transducer must be thin and robust and to be inserted with the sole of the shoe.

There have been several papers which have reported about the use of different techniques using insoles and their response to different compressive loadings. The different errors associated with each of these techniques have also been explained. Some of these techniques have been stated in chapter 2.

1.4 SOCKET MEASUREMENT.

Artificial sockets act as an interface between the human residual limb and prosthetic leg and should be designed properly to achieve stability and efficient control for mobility with smooth load transmission (Mak, Zhang, & Boone, 2001). Clinicians and biomechanical analysts have been very keen on understanding the load distribution of the limb to the load bearing areas within the socket. Therefore in order to design a better socket it is necessary to evaluate and understand the mechanical load distribution and how the limb responds to the external loads and at the limb-socket interface.

It is important to measure the pressure difference between the socket and stump for the evaluation of socket design in prosthesis. Pressure distribution and force application has an important role while considering the overall design of the prosthesis. The physician should have an idea about where the patient is applying pressure, the amount of force he applies along with the calibration of the system itself. If the physician is able to see the force distribution map or pattern, he can identify the spots where pressure is applied. For this, accuracy and precision plays a significant role in deciding the pressure readings. Accuracy can be defined as the per-cent error involved in measuring the applied pressure, while precision can be the amount of error variance. Hence, accuracy and precision is a key aspect when such data's are evaluated. It is made sure that the experiment reproduces the same results repeatedly, with 2-5% margin of error (Orlin & McPoil, 2000).

Pressure measurements in the sockets have been undergoing experiments since the past 50 years. This information has led to the understanding of socket – load transfer, assessing socket design or validation of computational modelling (Mak et al., 2001). Mak et al., (2001) studied that interfacial pressure measurements requires proper measurement technique, placement of sensor technique and related data acquisition. An ideal device should be able to measure the interfacial pressure and shear. Pressure measurement of the socket would be helpful if the readings were to match between the pressure magnitude and subjective sensations obtained by verbal reports (Neumann, Wong, & Drollinger, 2005). Mapping the pressure data within the socket would be of much more advantage if it assists in modifying the socket to the amputee's requirement and comfort.

The sensors for measuring interfacial pressure can be placed in the sockets in two ways. Either they can be placed in between the skin and the socket or by inserting them through the socket (Mak et al., 2001). With these methods it is not necessary to damage the sockets but because of their finite thickness, interference cannot be prevented because of the protrusions from the socket. The sensor size is also another factor to be considered. It has the similar conditions to that of insole force transducers. If the sensor size is large only the mean pressure would be measured and small size sensors would lead to 'edge' effects. Therefore it is necessary to have the sensor be precise in thickness. Due to the finite space between the limb and the inner surface of socket, placement and positioning of the sensor is essential. Improper placement of the sensor will show erroneous readings. Hence, the positioning the transducer has a major role while placing the sensor through the socket by making holes through them (F. Appoldt & Bennett, 1967).

Leavitt,et al (1972) conducted a study based on understanding the physiologic response of stump tissue to the socket interface and evaluated the forces acting on the tissue by the socket. They first made a record of the patients prosthetic history by interviewing him and later a video recording was taken to classify the amputees into ‘good’ and ‘poor’. A universal harness; consisting of cables, binary foot switches and goniometric modules is attached to the amputee’s leg. Later with the history data provided, pressure transducers are placed at certain anatomical positions. The patient is made to walk three times each and data from the ‘universal harness’ is send to the pre-conditioning circuits. This method is called as diaphragm deflection method and is used to analyse the gait analysis of an above knee amputees that with a normal individual (Leavitt et al., 1972).

Another type of measurement technique explained by (F. A. Appoldt, Bennett, & Contini, 1969) was to find the pressure irregularities when the pressure sensor protruded into the flesh of the patient. For this they used pressure sensors that was mounted onto socket and worn by the subject. The subject was then made to walk with the sensor on the flush condition of the skin followed by pressure test with the sensor protruding $1/16^{th}$ into the skin. The two flush gauge sensors tested here was the N.Y.U pressure transducer and the second flush transducer was Microsystems. For clinical tests the transducer was supplied by Sensotec Division of Scientific Advances, which had strain gauges on one side and a pressure sensing diaphragm on the other. It was found that the protruding gauge results were higher than the flush gauge results.

From the results obtained by (F. A. Appoldt et al., 1969) showed that protruding pressures were higher than the flush pressure. Pijkeren, Naeff, & Kwee, (1980) designed a new technique using hydraulic pressure system to measure the pressure applied at the socket interface which could measure pressure up to 700kPa applied at the interface. A small bag filled with silicone oil was inserted into the socket and as pressure was applied at the interface, the pressure change was detected by the pressure transducer.

Due to certain limitations such as, the need for designing special sockets and the fact that the sensors could measure only localized pressure (Engsberg, Springer, & Harder, 1992), used a printed circuit sensor mat that could measure the pressure at the socket interface. The map system developed at Tekscan was a thin pressure mat that could easily fit into the socket-limb interface without altering the socket design which could measure areas up to 25mm square region. Tests were conducted by making the patient wear the pressure mat between the residual limb and the socket and making him walk and stand while the pressure reading were recorded in the system. This kind of measurement was found valid, as it was able to overcome the limitations that were being raised initially.

Commercial systems currently available include the Tekscan **F-socket** pressure measurement system which is force sensing resistor transducer having 96 sensing elements covering an area of 155cm^2. This system can take up to 456 frames of data at a sampling frequency of maximum 165Hz till 700kPa of load. The advantages are that it is thin and flexible (made on Mylar substrate around 0.28mm thickness) and has a good sensitivity reliability and frequency response (Buis & Convery, 1997). The other available one is a sensor pad **Pliance** 16P system from Novel (Novel, 2004). The company stays on using the capacitance sensors in a 4X4 matrix cell. The 16 sensors in it can be used to monitor different anatomical areas of the limb simultaneously with pressure ranging from 20–600kPa. The dimension of the pad conforms perfectly to the anatomical areas of the limb. The final one is the **Rincoe Socket Fitting System** which has 60 sensing cells arranged on a 0.36mm thick polyvenilidyne fluoride strip. The sampling rate of Rincoe system is 100Hz and can measure load limit up to 83kPa. The polyvenilidyne substrate is brittle and does not conform to the contours of the limb and caused damage to the sensor pad. Hysteresis, drift and accuracy errors were quite high for Rincoe system when compared to Tekscan’s and Novel sensing system (Polliack et al., 1999)

Obtaining pressure from an amputee to design a trans-femoral and trans-tibial socket has been a challenging task. The maximum and minimum pressure readings have been found to be 59.33 kPa at the distal femur and 24.8 kPa at the lateral proximal brim.(Neumann et al., 2005) The designing and the fitting process requires collecting data from the patient. It can either be personally asking them about what he/she is experiencing, by visually observing the contours of the amputee leg and observing the discoloration of the skin or by touching the bony projections. There have been clinical tools that have improved the quality of data for the better understanding of designing the sockets. But along with it comes glitches in the system which needs more reviewing.

1.5 AIMS AND OBJECTIVES

Aim of this project would be to develop a transducer for the measurement of foot and socket pressure. The current product being designed here should be able to overcome the errors found in previous system and provide better pressure readings with precision and accuracy. The objective here is to design a transducer which should be as small as possible so that it can fit into the sole of the shoe allowing a uniaxial compression of 0.5 mm. A new technique using optoelectronics has been incorporated to measure the intensity of load being compressed. The compression activity is facilitated with the help of a rubber. The criterion under which the rubber is chosen depends on the mechanical properties of the rubber such as strength, compression factor, loading responses etcetera. The following chapters present a detailed discussion about the design criteria.

2 PLANTAR PRESSURE MEASUREMENT

2.1 LITERATURE REVIEW

Plantar pressure measurement uses discrete transducers and today many commercial systems have come up for this type of measurement. Different companies have developed transducers either using the discrete measurement where the sensors are positioned at specific anatomical points (figure 1.1 and 1.2) under the plantar surface of the foot or the matrix measurements where an array of sensors are organized in rows or columns which evaluate the entire planar surface of the foot (Orlin & McPoil, 2000). Discrete measurement system uses less number of transducers and hence can be used for motion analysis like running or any sports activity. Since the number of sensors used is less, the clinician has to decide the proper positions for placing the sensors under the foot. These sensors are then attached to the foot either by the adhesive tape or by embedding them into an in-sole and then placing it in the shoe. The advantage of using a discrete measurement technique is that the sampling frequency can be high up to 200 Hertz and hence a large number of readings can be taken. Matrix measurement system uses larger number of sensors and hence can assess pressure at once (Orlin & McPoil, 2000). The main disadvantage of using the discrete system is that it can have ‘edge’ effect due to inconsistency in the position of the sensors when placed on the insole. Furthermore, since the sensors are either taped on or embedded into the sole, the sensors can move around and that can change the desired anatomical points specified earlier. In the case of the matrix measurement, a prior assessment is not required and can map the pressure over a large area.

Having done their research, Orlin & McPoil, (2000) showed how plantar pressure measurement be utilised for assessment and for the understanding of lower foot disorders associated with neurological and musculoskeletal systems. Mueller K, Cornwall M W, (1995) investigated the effect of tone-inhibiting dynamic ankle-foot orthoses on the foot-loading patterns of patients with hemiplegia. The results showed that the ankle-foot orthoses increased force and impulse through the foot during contact with the supporting surface. Researchers have since then assessed various insole materials and the shoe modifications for treating foot disorders.

Billing, (n.d.) mentioned about the advancements made to the insole in measuring the kinetic variables by pattern recognition analysis. He experimented using PVDF (polyvinylidene fluoride) electromechanical transducer as it was thin and also had high output voltage, had high mechanical strength and that could be molded to different shapes. However, it was subjected to lateral and axial motions apart from the perpendicular movements that bought irregularities to the readings.

Some of the discrete pressure transducers currently used are *F-Scan System*, manufactured by Tekscan, *Pedar-insole system* by Novel Electronic, *Musgrave footprint system* by WM Automation and Preston Communication. All these systems have different technology and the measurement technique unique to each system. Plantar pressure can be measured using either Force Sensitive Resistance (FSR), or by capacitance method, microcapsules, MEMS (Micro Electro-Mechanical Sensors) or even load cells. However, only FSR (force sensitive resistors) and capacitance technologies are widely employed for measuring the plantar foot pressure .

FSCAN™ is a very thin-layered transducer using force sensitive resistor technology with metal patterns printed on two Mylar sheets with a conductive polymer layer embedded between the two sheets. The inner surface of one sheet gives a row layout, while the other sheet provides a column layout This assembly of row and columns creates a sensing cell or ***sensel***™ (Tekscan, 2007). There are about 960 sensors on a disposable 0.1mm thick insole. Thin aluminium tracks are deposited on the external surface of the sheet to provide connections to instrumentation unit worn on the ankle (SENSORS, 1991). As the contact area increases, the resistance between the conductive sheets decreases as the pressure between the Mylar sheets increases(Cavanagh, Hewitt Jr., & Perry, 1992). Hence change in resistance causes a voltage change in the system, which is recorded and displayed. FSCAN sensors are can be trimmed in specific dimensions according to the user's needs. However, a study on FSR by (Nicolopoulos, Anderson, Solomonidis, & Giannoudis, 2000) showed that due to the electromechanical problems, large errors were produced on the output results thus raising issues on the reliability of the system. The output of the system was affected by poor hysteresis, inaccurate calibration, shear and temperature, thus reducing the durability of FSCAN^{TM} system.

Another commercially available method for measuring plantar pressure is the Musgrave Footprint system which uses force-sensing resistors It consists of two polymeric sheets, one coated with pectinate electrodes and the other with a semiconducting (molybdenum disulphide) material. The working principle is the same as it is mentioned above. The thickness of the device is 0.25 mm – 0.7mm thick. The sensors are within the measurement range of 0-4MPa per sensor while the system works in the range between 11 – 110kPa. The sensors responses changes by ±2 % when the load is above 11kPa and by ±15% for loads above 1MPa (Cobb & Claremont, 1995).

It was Nicol & Henning, (1976), who described the use of capacitance transducer for plantar pressure measurement. Capacitance transducer uses two plates of a conducting material separated by a di-electric material. When force is applied, the distance between the dielectric decrease and that causes a change in the capacitance. As the distance decreases, the capacitance increases and the resulting voltage change in noted. Novel Electronic incorporated this technology in their Pedar insole system. They incorporated around 99 sensors in 2mm thickness insole with a measurement range of 30kPa to 0.6MPa with 1kPa sensitivity. Even with bending forces, the operation of the device was maintained.(Cobb & Claremont, 1995) But it was the size of these sensors (2mm in thickness) that made it less competitive than the other sensors. Moreover, it was expensive to develop this kind of technology as it needed more charge amplifiers.

Henning fabricated a shoe insole using 499 piezoelectric ceramic (lead-zirconate-titanate) transducers of dimensions 4-7 mm^2 X 1.2 mm that was embedded in silicon rubber. With a sensitivity of 0.5kPa, the range of measurement was 0-1.5MPa with hysteresis as low as 1%. Due to the variation in sensitivity because of the material and fabrication tolerances, that was necessary enough to calibrate each of the sensors. The main sources of errors were considered to be of (i) pyro electric charge generation, (ii) susceptibility to electrical interference, and (iii) sensitivity due to lateral strain. The other problems related to these types of transducers are that they are difficult to be constructed and also can undergo mechanical fatigue.

Micro Electro-Mechanical Sensors (MEMS) was another technology proposed by Wahab & Bakar, (n.d.) suggested the use of MEMS as a latest technology integrated with biomedical instrumentation to achieve real time and efficient measurement. The main advantages of MEMS were the miniaturization in size, low power consumption and integrating the sensors respect to control circuitry. For foot pressure measurement, piezoelectric MEMS sensor was incorporated for the experiment. The mechanical setup was mathematically modelled for membrane thickness and side length with silicon as the material. It was finalized to a square shape design as per Von Mises stress testing. Once the custom-made mechanical setup was done with, pressure testing was done for the calibration. Pressure was applied from 20 psi to 135 psi. The research was successful as they were able to prove the linear relationship between pressure and the output voltage.

Chedevergne et al (2007) had proposed a new method of analysing plantar pressure using piezo electric sensors under barefoot conditions. They chose to install their device in a shoe without any insole, because they felt that insoles constrained the foot and the dynamic contact to ground. The sensor is designed using dynamometric rings held in position by metal cases and further sandwiched by aluminium plates. So when the subject walks with the shoes on, he compresses the force sensor. This compression causes a change in voltage, which is amplified before acting as input to a personal digital assistant (PDA). This data can be then transferred to computer to plot the pressure points and data. Static calibration was done using INSTRON 6025, by applying compressive loads from 0-2000 and back to 0N at frequency of 100Hz lasting for 24seconds. They were able to study vertical forces and also estimate the velocity of centre of pressure, duration of the stance phase, size of contact area at a certain time, maximum loading rate et cetera. Due to the shape of the sole, the user did not feel any disturbances. It offered linearity precision and longevity. Also walking barefoot, the senor was able to give in plantar force and pressure between the foot and the ground, with local pressures helping in analysis.

Work done by (Hennig & Milani, 1995) illustrated that in-shoe pressure data offered more information about the loading behaviour of the foot. They were able to compare the characteristics between the shoe and the foot complaints. Henning and Milani had used eight piezo-ceramic transducers, placed at specific anatomical positions, for their study and analysed the loading patterns for 22 different subjects. The subjects were asked to run across a force platform and the pressure readings were noted. The force curve and the summed pressure signal were time normalized using an interpolation technique (Cavanagh P.R & Lafortune M.A, 1980). The vertical GRF had an initial impact of 1.69 BW (body weight) followed by a maximum force of 2.63 BW. Higher lateral pressures than the medial pressures were noticed during foot-strike and the eight anatomical positions were finalized as the main load bearing positions.

Chesnin, Selby-Silverstein, & Besser (2000) conducted studies where they compared an in-shoe pressure measurement system and a force plate for validating the centre of pressure. Centre of pressure (CoP) was defined as the origin of the GRF while walking. The in-shoe system used here was by Parotec System compared to an Advanced Mechanical Technology Inc. force plate. Parotec system consisted of a microsensor placed beneath a hydrocell that could be deflected only at the top and bottom. Centre of pressure was calculated and it was found that the correlation coefficient from the two systems were greater than 0.70 for 52/67 trials in the ML direction and were greater than 0.90 for 67/67 trials. The mean error was 1.37±0.59 cm showing that the Parotec system was better when compared to AMTI force plate system. Though there were slight errors, it could be neglected, as it was inferred that it could be the placement or the movement of the sensors inside the socks was the reason for errors.

Critical light reflection is another technique used for plantar pressure measurement. It uses the principle of light reflection as the basic technology and has a video camera and micro-computer for data storage. An illuminated rectangular glass plate fitted with force transducers and as force is applied, the intensity coming out of the glass is changed and that intensity is recorded by the video camera. The force transducer measures the magnitude of force and with the intensity of light detected, the computer analyses the amount of plantar pressure (Orlin & McPoil, 2000). Commercially, some of these systems are referred as Pedobarographs. It consists of a glass plate, which is covered by a thin sheet of plastic or rubber. Light is internally reflected from the sides into the gap and follows the critical angle of reflection theory. So if the angle of incidence is less than the critical angle of reflection, the light gets transmitted through the surface when pressure is applied from the top and this is the working principle of pedobarograph. This maps the pressure and different colours are assigned with respect to the intensity of the pressure applied.

Vertical forces have undergone thorough studies in the past few years. Some of the currently available ones have been mentioned in the preceding paragraphs. But Hughes, et al, (2000) found that these systems needed calibration which otherwise would affect the reliability and the resolution of the pressure values. The study also showed that smaller sensor size also recorded larger pressure, but at the sacrifice of resolution and cost. So therefore Hughes, et al (2000) developed another technology based on interferometry technique for the development for the high resolution foot pressure measurement. Basic principle employed here is the use interferometry technique which detects the compression of the pressure plate (Perspex) by a load (up to 15kg) using a He-Ne (633nm) laser light. The pressure distribution is represented by an interferogram (interference pattern) which is given as an input to a video camera and later processed to display a three dimensional map of the pressure distribution. The advantages of considering this system over the commercially available ones are the high resolution and the low cost of manufacturing this technology. The resolution of the system is found to be 70 pixels per cm^2 and could measure up to 1-1000kPa.

Another commercially available in-sole pressure transducer is the hydrocell that consists of a piezoresistive sensor encapsulated in a fluid cell which is then embedded into an insole. So as the pressure is applied to the hydrocell, the resistance with the water increases and the piezoresistive sensor detects and produces an electric signal proportional to the resistance change. This technology has been employed in Parotec system, (Paromed Medizintechnik GmbH, Germany.)

Medilogic^{TM} (Medilogic, 2012) has manufactured an insole used to measure foot pressure. Covering a range from 0.6 – 64 N/m^2 it records readings at a sampling frequency of 60Hz. It comprises of two thin in-sole with 200 sensors that can be shaped to the customer's needs. A wireless modem helps to send the data instantly to the computer without causing any hindrance to the patient. Recording of the data can be done on well organised Medilogic® software which has the provision to record and stop the recording, also to present the data in an isobaric colour display along with numbers on a particular position. Data can be saved in a hard disk for further evaluations. They also have prosthesis socket measurement with six freely position able strips each having around 10 sensors each (Medilogic, 2012).

2.2 COMPARISON BETWEEN INSHOE AND PLATFORM

Ever since the purpose of clinical evaluation has been considered important for gait analysis, the manufacturers of force measurement systems have found it necessary to develop pressure measurement techniques on a larger scale. Sooner, the companies started to develop the systems for the clinical purposes, but there were certain issues that had to be considered when selecting a specific system. The number of sensors used which provided higher resolution and the positioning of the sensor parallel to the plantar surface to give a vertical force measurement were few of the issues.

In matrix measurement system, the number of sensors used is less than the discrete system and hence there can be reduction in the resolution of the system. Furthermore, due to the shear forces acting on the insole, the probability of the individual sensors getting displaced within the insole is high. Moreover continuous mechanical loading can cause damage to sensors. Extra care has to be taken when pulling out the connection cables, as they may have the chance to bend or break (Orlin & McPoil, 2000).

Jean-Pierre Wilssens of RSScan International has stated that the company uses their own patented foot-scan system which uses resistive technology to measure foot pressure. It has been used for diagnosing pathologies and also medical interventions. There have been studies which have shown that foot-scan systems have been used to assess the morphology and functionality and the effect of footwear on the foot. These systems are found to be efficient when working at higher frequencies of 500Hz. The company claims that the measurements are taken at higher frequency when the foot is stimulated to its dynamic range of loading.

Since discrete and matrix measurement technique systems have their own advantages and disadvantages it is up to the clinician to make a selection on the system he would like to use for assessing his patients whilst depending on the nature of the activity (Orlin & McPoil, 2000).

By this experiment, we aim to measure the plantar pressure using in-shoe method. The socket pressure measurement will not be feasible at this stage of the project. Hence, initial trials will be conducted by applying static compressive loads to the device. An electromechanical model is setup using the optical pressure transducer. Forces applied to the transducer will be replicating the pressure applied by the foot and is designed by considering the literature review and its possible future opportunities.

3 CURRENT TECHNOLOGY

3.1 PRESENT DAY SYSTEMS

Resistive and capacitive sensors are the most commonly used plantar pressure sensors. Most of the resistive sensors have the same principle. They undergo a change in the electric current flow relative to the pressure exerted on the sensor (Superiore & Giacomozzi, 2010). Force-Sensitive-Resistors, one common type of resistive sensors, works on the principle that small scale deflections causes an effective increase in the contact area thereby producing a variation in the electrical conductivity which linearly resembles the pressure applied (Superiore & Giacomozzi, 2010). This kind of sensor falls under the 'surface effect' sensor category. The other kind of sensor based on the resistive principle is the 'volume effect' sensors where the change in volume due to elastic deformation is evaluated. Volume effect sensors have conductive particles that are dispersed in a polymeric matrix which when deformed causes change in the volume. The resistance of these sensors are non-linear and actually requires low voltages for operating them in linear region. Since the sensor has low impedance, there are not much of the noise effects on the measurements.

An in-shoe pressure measurement system determines the pressure distribution of the foot. This has been very much used for clinical applications for understanding the gait movements and analysing the pressure patterns that helps to evaluate the gait control. There are different pressure sensors available, but when it comes to plantar pressure sensors, the main ones fall into the 'resistive' and 'capacitive' technology. The basic working principles of the two technologies depend on their general electrical working.

Researchers in *AMCUBE* designed a capacitive pressure sensor mat for the diagnosis, management and for scientific research in biomechanics. They were able to construct a sensor plate with 2 capacitive sensors of square centimetre and an active area of 49 X 49cm X 5mm. It could be used in both static and dynamic gait analysis conditions. For the static analysis, pressure loads were applied in the range

2-100 kPa and for dynamic conditions, the pressure values went up to 900 kPa. They chose capacitive sensors over the resistive and piezo-electric types, owing to the fact that capacitive sensors gave better pressure readings with absolute accuracy and worked over a wide range of frequencies. Being light weight and cheapest in the market were the few advantages the device had. Since the connectivity was by a single USB cable, it was easily portable also. Moreover the device had to be calibrated only once which showed its efficiency and even after repeated loadings, the measured values never deterred from the normal range (Superiore & Giacomozzi, 2010).

Novel believed that accurate and reliable information was the key to clinician's appropriate treatments for gait irregularities. In certain pathologic conditions such as diabetic foot, hind foot valgus deformity et cetera, the pressure readings are extremely important. As the loading pattern of the foot changes and also the local pressure values change drastically, some reaching up to 12MPa. Hence for better accuracy *Novel* designed a pressure device, EMED pressure platforms using capacitive technology. The system included highly calibrated accurate capacitive sensors and a spring element to balance the deformation. The mechanical deformation is then translated into an electric signal. By using spring balance element, the system is able to reproduce the deformation over a long period of time. For calibrating procedures *NOVEL*, used Trublu calibration system that was able to correlate the applied pressure and the displayed sensor signal. The accuracy was found to be +/-5 % for pressure ranges up to 1.25MPa. However it was suggested that the calibration process had to be carried out periodically as the sensor properties were subject to change over the period of time. (Superiore & Giacomozzi, 2010)

3.2 PEDAR AND FSCAN SYSTEMS

The recent advancement and needs of clinical trials and evaluation of the gait analysis has led to the further validation for accuracy and precision of the insole plantar pressure measurement. Hsiao et al, 2002, conducted experiments on the FSCAN and PEDAR insole systems and they were able to find that the PEDAR system was more accurate and precise when it came to pressure applied, calibration procedure and also on the duration of the applied pressure. (Hsiao, Hongwei, Guan, 2002)

Antonello Fadda, (Superiore & Giacomozzi, 2010) has given a basic idea about capacitive sensors The working of capacitive sensors is based on the change in the thickness of the dielectric material compressed between two metal plates. As pressure is applied, the dielectric material gets closer to each other and since the capacitance is inversely proportional to the thickness, an increase in pressure produces a proportional change in the capacitance value. However there are couple of drawbacks when compared with resistive sensors. Capacitive sensors can only be used for slow responsive input signals. Also small capacitances are high impedance devices and should be carefully designed in to prevent noise and interference problems.

3.2.1 PEDAR

Arndt, A (2003) conducted test on Pedar matrix insole and found a 17% creep after testing for 3-hour on the Pedar matrix insoles. Moreover it was reported that discrete insole devices had disadvantages when compared to the matrix insole devices (Abu-Faraj, Harris, Abler, & Wertsch, 1997) which was mostly because of the dis-positioning of the sensor and also the transducer might act as foreign body. There was not much of the literary work that was supporting the validity of theses insoles when assessing the vertical ground reaction forces.

Hurkmans et al., (2006) found that Arndt method of validating Pedar was not optimal because the patients were using walkers and hence the system could not measure the total body weight. So Huckmans et al. did a validation experiment on Pedar for 7h. They used Pedar mobile system, which had 99 capacitive sensors in the insole to measure the vertical force. Pedar insoles were calibrated using Trublu calibration device from Novel. Simultaneously vertical ground reaction forces were also measured using Kistler force plate. Forces were applied from $4N/m^2$ to $60N/m^2$ with a sampling frequency of 99Hz. Data samples from the force plate were acquired at 500HZ. The offset drift found in Arndt report was limited by using a custom made correction algorithm. Both static and dynamic testing was done on the Pedar system.

The study showed that there was a 12% drift error which was present after 4 hours. Further assessment into the Pedar data showed that there was a 14 % drift after the 7h testing, even though the drift was insignificant in the first 3h. The difference in drift was dissimilar for different patients. The author assumed that this might be due to the fact that the insoles would have been old and this statement was proved by (Hsiao, Hongwei;, Guan, 2002). Data from static and dynamic loading showed that the offset difference was the same for both and remained a constant for the 7h period.

Figure 3.1 PEDAR
http://www2.hud.ac.uk/hhs/
chsup/equip/pedar.php

3.2.2 FSCAN

It is necessary to calibrate any system prior to using them for evaluation purposes. This reduces the possibility of hysteresis and creep on the following tests (Flórez & Velásquez, 2010). For static and dynamic calibration of the sensor, a FSR was connected to a voltage to current convertor and the data acquisition is carried out by Labview, (National Instruments). The compressive load is controlled with the help of a nut and screw mechanism which moves the load cell up and down.

Static testing was done by placing calibrated weights over the sensor, to obtain a relationship between the applied load and the output voltage. Voltage values were collected from the system by implementing an algorithm in Labview. Dynamic testing included the loading and unloading on the sensor. From both the tests, certain characteristics were observed which was undesirable to the sensor. Static measurements showed effects of creep and dynamic characteristics had hysteresis which was later compensated by processing the acquired data. The authors concluded that calibration is an important aspect when it comes to measuring low pressure range (0-4N) because that it where most of the non-linearity happens and it is necessary to calibrate each individual sensor as they differ from behaviour differs from one another (Flórez & Velásquez, 2010).

Nicolopoulos et al (2000), reported that FSCAN had irregularities in the electromechanical aspect of the sensor which started to show erroneous readings. The overall accuracy of the instrument was depended on the inaccurate calibration, hysteresis, bending and shear effects on the output of the system. About 10-20% of hysteresis was observed when the system was loaded below 50 pounds and for maximum loading (Crammer & Patterson, 1992). The same was also reported by (Rose, Fewell, & Cracchiolo, 1992)but only that these sensors were subjected to experiments on different days (Nicolopoulos et al., 2000).

For the calibration purposes, the subjects were asked to stand on the insole with a known load applied to the insole. The calibration process was either a small volume calibration or a large volume calibration. For small volume calibration, the FSCAN system was initially loaded up to 80 N, with incremental values of 10 N. The samples were taken at a frequency rate of 50 Hz. For larger volume testing, the INSTRON loading machine was used. Both static and dynamic tests were done with the FSCAN placed in between a metal block, indenting the shape of the foot and a bottom plate. Compressive load up to 0.8000N is applied to the FSCAN area at about 530 kPa, which is above the mean peak pressure of 140 kPa under normal standing pressure (Cavanagh Peter R, Hewitt FG, 1992). To test the dynamic responses of the sensor such as hysteresis, repeatability and the frequency response, a triangular input load cycle was applied till maximum load of 4000 N at a frequency of 0.1 to 0.5 Hz. However the system failed to produce frequencies above 1 Hz, which is the normal frequency of the human gait. Also with repetitive dynamic loading, the two sheets of FSCAN started getting close to each other as an effect of continuous loading, hence decreasing the resistance and thereby increasing the output forces. One major concern related using FSCAN was its incredibility to record accurate readings under bend surfaces. (Nicolopoulos et al.2000).

Figure 3.2 FSCAN
www.btsbioengineering.com/BTSBioengineering/pressureanalysis/BTSFscan/BTS_FScan.html

4 FUNDAMENTALS

4.1 FORCE SENSORS

Force sensors are mechanical sensors that measure the intensity of force applied. These kind of sensors should be small, thin and flexible and be detecting in the range of several hundred Newton. The sensor should be fitting in to the contours of the shape it need to measure (Lai & Li-Tsang, 2009). The force sensing device will respond to basically two types of forces (shear and vertical forces) irrespective of the area applied or the point of application (Urry, 1999). These conditions are usually satisfied if the contact surface of the sensor is relatively stiff with the skin, which is usually from aluminium or steel. The sensor output can vary depending on the intensity of load applied and hence can operate unpredictably. An ideal sensor should give an output inversely proportional to the area, under constant force conditions (Urry, 1999).

Any force sensors should be calibrated prior to their use. Static and dynamic calibrations are the two methods by which the force sensors are usually standardized. Static calibration is done by applying a known load and then checking the response of the sensor. The response changes depending on the materials used and can have undesirable characteristics such as creep and hysteresis (Pitie D, Ison K, Edmonds M E, & Lord M, 1996). Testing for dynamic calibration is carried out by loading and unloading the sensor within a specific time interval using testing equipment (Dhanendran, Hutton W, & Paker, 1978). Force sensors are usually calibrated and are considered to be the same each time when a load is applied through some kind of rigid structure. If the load is applied through some fluid filled bladder or membrane, the output is usually pressure and is classified to be a pressure sensor (Urry, 1999).

The size of the force sensor is another factor which needs to be considered. Transducers with smaller size sensors placed over a region of high pressure can record good pressure readings, while the larger ones on the same spot can give in a mean pressure value surrounding that area. But these pressure readings will be an estimated value but lesser than the actual quantity. Therefore the positioning and the placement of the sensor are important for true and accurate readings. So for taking readings, an active surface area is of 5mm X 5mm is specified. The frequency response of the transducer should be similar to that of the frequency of walking. Reports showed that the normal walking frequency was between 10 Hz and went up to 20 Hz (Antonsson E & Mann R, 1985), (Giakas & Baltzopoulos, 1997). Hence for the current experiments, 10-15 Hz was considered as a minimal range of frequency and for responses under the heel, frequencies up to 200 Hz was considered (Nevill, Pepper, & Whiting, 1995). Duckworth et al, (1982) reported that pressure ranges measured during stance and during walking were 0-200 kPa and up to 1000 kPa respectively.

Output from the force sensors are mostly electric signals. These signals are favourable, as it can be processed, stored and analysed later. While for some systems, the output taken is in the video format. One such format is the optical pedobarograph, where the video of the pressure compression was recorded in an optical method (Duckworth et al., 1982).

4.2 DESIGN CRITERIA.

The mechanical design of the pressure transducer has been shown in fig. 4.2 to fig. 4.10. The mechanical design of this transducer has been designed taking the general characteristics of a foot pressure transducer.

1) Small and robust in design as well as in size.
2) Strong enough to withstand any pressure applied.
3) Detect the minute pressure changes.
4) Miniaturized to suit any surface.
5) Comfortable to the user.

A detailed description about the design of the transducer has been mentioned in the following paragraphs.

1) Size: Since the concept of designing a new transducer is to measure foot pressure from within the sole, the size of the transducer is a critical criterion which has to be of main concern. Initial design of the transducer will have

dimensions of diameter 28mm and height 11mm. The height of the rubber used here in 6.5 mm, which gives a total height of 17.5 mm. These dimensions are considered for initial testing to prove its feasibility using optoelectronic technology.

2) Weight: The weight of the device is another important criterion to be noted. The device should not be heavy and cause any inconvenience to the wearer. The lighter the device is, the better it will be. Currently, the objective is to manufacture the device that weighs around 50 grams and then decrease it according to the specifications needed.

3) Stiffness: When considering the compression of the device, stiffness has a vital functionality in the designing of the device. The device should not be too stiff nor be too soft. The stiffness should be corresponding to the intensity of the load applied.

4) Maximum load: The device is designed to endure a maximum load of 800 N. from a no load condition.

5) Maximum deflection: The output of the transducer is measured in voltage which is corresponding to the intensity of rays detected after deflection from the surface. Hence by compressing the piston, the angle of deflection changes and this further induces change in the intensity of light detected by the phototransistor. As the compression increases, the angle of deflection also increases and a maximum angle of

6) Frequency range: A minimal sampling frequency range of 10-15 Hz is selected for taking the output readings from the system.

7) Thermal response: The temperature range in which the device works is chosen to be between -5°C to 30°C, which is the normal room temperature.

8) Uniaxial compression: The transducer designed here is being tested to measure the intensity of uniaxial applied to compress the rubber. The main aim is to attain a compression of 0.5 mm for the intensity of load applied till the output reaches the saturation level.

Various force sensors used have different mechanisms depending on the application. Most of the sensors detect motion and hence respond to six degrees of motion. The six degrees of freedom (DoF) represents the motion of an object or a rigid body in space in three dimensions and it includes a set of translation motion in the 3 perpendicular axes and rotational motion about the 3 axes. The translational motion in the three perpendicular axes includes displacement in *forward/backward, up/down, left/right* direction. The rotational motion around the axes includes *pitching*, *yawing and rolling* represented as in figure 4.1.

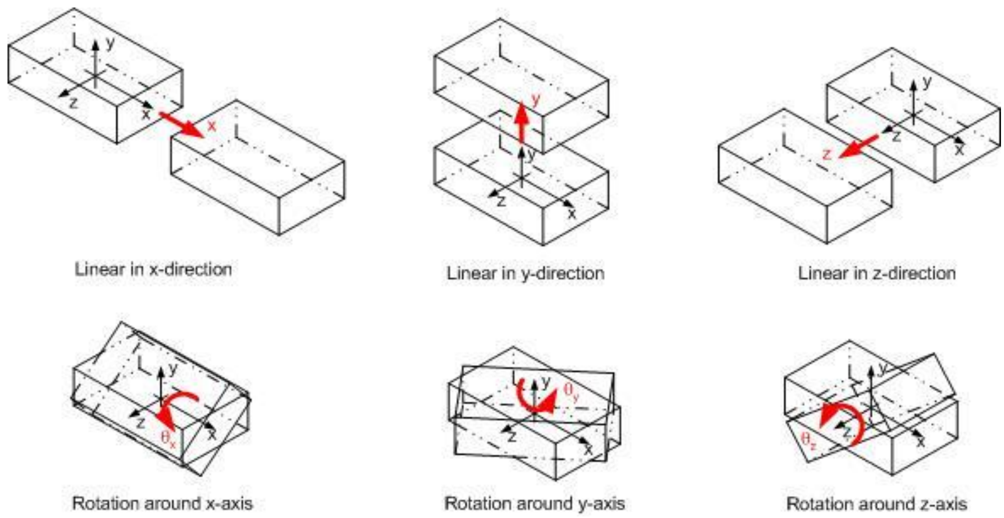

Figure 4.1 Six degrees of motion
http://www.oring.com.pl/oferta/f1-simulator-3-dof-motion-platform

The main aspect of designing a single axis force transducer is to avoid *5-DoF* and focus onto the displacement in the *y-direction.* The main motive of the mechanism of the sensor is to detect displacement only in *y-direction* and exclude displacements in other directions. Hence the main characteristic of designing the sensor is to allow only compression in one dimension.

For the initial testing, the sensor is manufactured using aluminium. Several materials were considered which included acetal bar, hardened steel et cetera. However, aluminium was considered as a suitable material for the casing of the sensor considering the availability and the easiness to use to any desired shape. Other attractive features included strength, resistance to corrosion, reflectivity, ductile and malleability.

1) Weight: Since weight of the whole device is crucial here, aluminium is considered for the main body. It is a light weight metal with a weight of 2.7 g/cm^3 which is considered to be $1/3^{rd}$ of that of steel.

2) Strength: With a tensile strength of 70-700MPa, aluminium is considered as a strong metal and hence it can be used to manufacture a device having a rigid body. Hence it should be able to with stand the maximum load applied to the device.

3) Machining: Since the device is small, extra care has to be taken to manufacture the device precisely. Aluminium can be easily be worked upon using machining tools such as drills, saw etcetera and the process of cutting, milling, punching, bending can be easily done with a low energy input.

4) Joining: If different pieces are manufactured separately with idea of joining them later, then aluminium can be joined using adhesive bonding or riveting.

5) Non-magnetic: Aluminium is considered as a non-magnetic material and hence can be used in presence of surrounding magnetic fields.

6) Reflectivity: When highly polished, aluminium exhibits a very good reflective surface. The basic working principle of the device is the reflection of light and it has to be made sure that light does not get diffracted due to the irregular surface.

7) Ductility: Aluminium is considered to be very ductile and malleable and its ductility allows products of aluminium formed close to the product design. So it can be designed to the user's requirements.

8) Opaque: Its feature to block off light form the surroundings helps this particular sensor to be used without any interference from the surrounding ambient light and thus help the photo transistor to detect the infra-red rays.

The sensor is designed to be circular in shape (figure 4.2). For specific reason, the circular shape is considered over square or rectangle. Initial designs were drawn with respect to square shape. A square *box-within-a-box* design was primarily considered owing to the reason that this design could sustain the basic necessity of the transducer design, which was to prevent *5*-degrees of range of motion. But when considering the main areas of pressure application, the contours of the foot are found to be circular in shape. For the ease of manufacturing and machining of the initial conceptual design and the fact that circular contours of foot, circular *box-within-a-box* shape is preferred over square design.

Figure 4.2 Piston and Shell

Figure 4.2 shows the basic design of the transducer. It consists of the piston and a shell that forms the pressure transducer and further description of the parts are described in the following paragraphs.

The transducer consists of three parts. The top part of the transducer is called the piston (figure 4.3 - 4.4) and is T shaped. The upper surface of the piston comes in contact with the foot. The piston consists of two components. The head with diameter 28mm and height of 4mm and the shank of diameter 17mm and height 11.5mm. The bottom part of the shank is a highly polished reflective surface for the rays to get deflected.

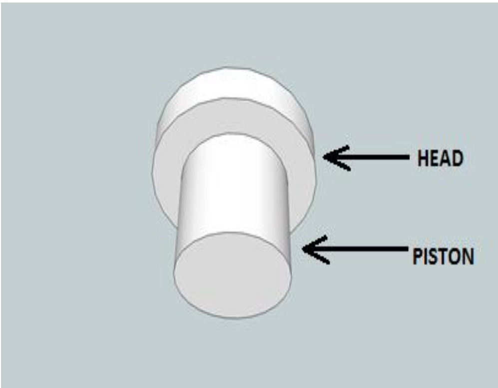

Figure 4.4 Parts of piston

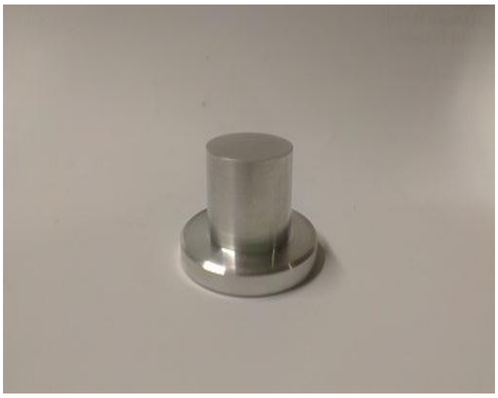

Figure 4.3 Mechanical piston

The second piece is called as the shell (Fig. 4.5 – 4.6) with dimensions of outer diameter 28mm and a thickness of 4mm. It has a slot from the top for the piston to move in the vertical direction. The inner diameter of the slot is of 20mm with a depth of 7mm, open at the bottom. The bottom face of the shell has an opening for inserting the optoelectronic circuit board. The diameter of this opening face is 17mm with a height of 4mm. A small recess sticks out where the circuit board rests on the recess. The protruding length of the recess is about 2mm and around 1.5mm in height. The height is just made so that it in level with the LED's. The height of the slot is 2mm more than the piston, so that a minimum gap is maintained for proper reflection. Figure 4.7 shows the internal front view of the device with slot and pit.

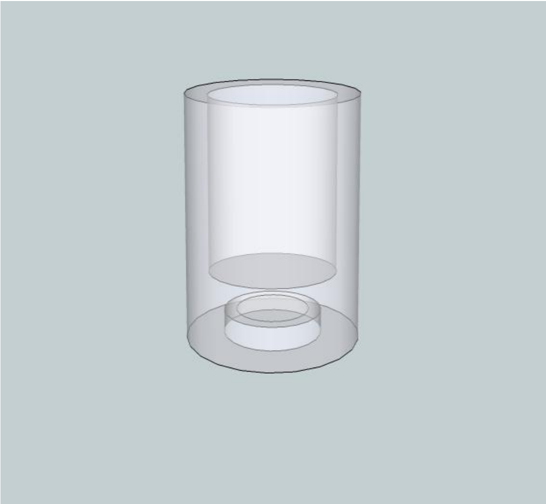

Figure 4.6 Inner structure of shell

Figure 4.5 Mechanical Shell with rubber

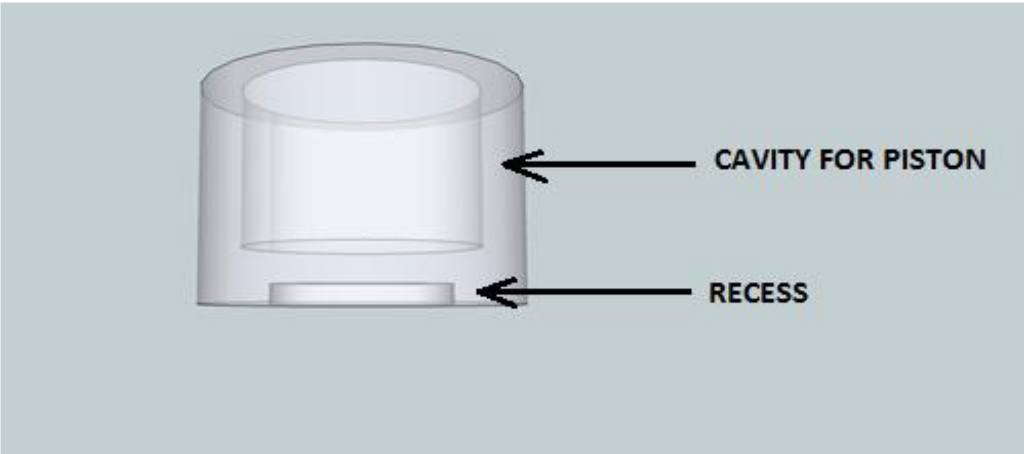

Figure 4.7 Parts of shell

The third part is the bottom plate which acts as closure for the shell. Apart from closing the shell it acts as a support and has four threaded holes through which screws are inserted. Two of these holes are for fixing the plate to the shell and the other two are adjustment screws which hold the circuit board in place to the recess. Placement of the holes can be seen from the fig. 4.8. The fixation screws (screw hole 1) are of 3-4mm in diameter and the adjustment screws (screw hole 2) are 1-1.5mm in diameter. The fig. 4.9 shows the bottom view of the position of the screw slot and figure 4.10 gives an idea about the thickness of the bottom plate.

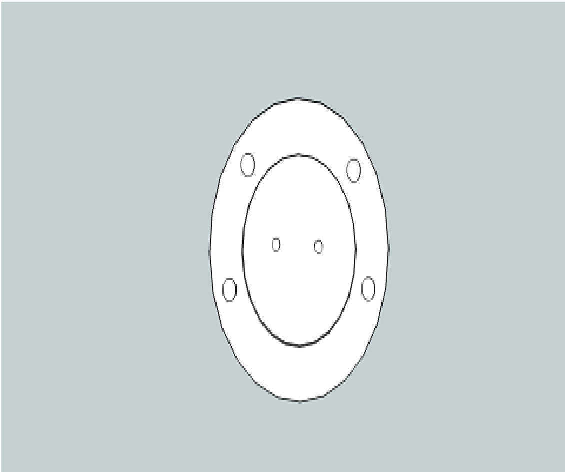

Figure 4.9 Structure of bottom plate

Figure 4.8 Mechanical bottom plate

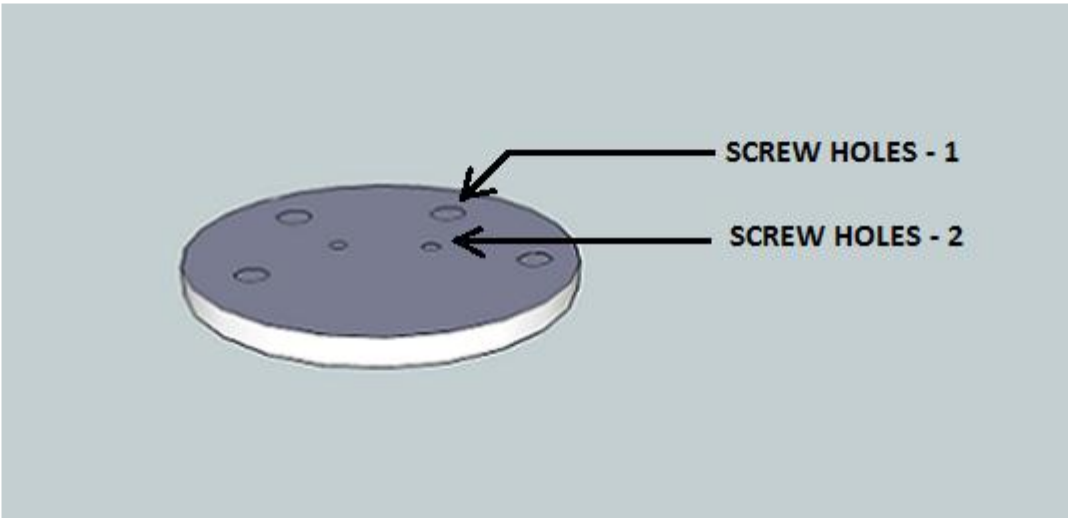

Figure 4.10 Parts of the bottom plate

4.3 RUBBER

An integral part of the transducer is the rubber. Rubber is chosen because of its property to compress under loading conditions and return to its normal state after releasing the load. A thin sheet of rubber with thickness ranging from 6mm to 1mm is sandwiched between the piston and the shell. The mechanical working of the transducer basically depends on the compression of the rubber. Load is applied on the piston which in-turn compresses the rubber leading to a deflection in the output.

Once the force on the rubber is released, the rubber decompresses back to its original shape. To choose a rubber, it is necessary to consider the fact that it has a good resilient rebound, so that once compressed the rubber can return to its natural position. The main characteristics of rubber has been stated by (Williams, Porter, & Roberts, 1992), stating that the assumed rubber should have high rebound resilience and low compression set and consequently low hysteresis. Bearing in mind about the type of experiment to be conducted, the stated characteristics are important to consider. Tests were conducted by (Tappin, Pollard, & Beckett, 1980) for measuring stress by the foot, where they designed a transducer using silicon rubber as intermediate device between the two plates.

The rubber that has been used for this experiment is industrial rubber used for applications such as providing cushion to the metal parts, acting as a shock absorber. The rubber is chosen as it should be an intermediate between being a soft rubber and a hard rubber. If soft rubbers are used, then the force acting on the transducer can compress the rubber so much that there is a possibility that the piston will be in contact with the diodes. Hard rubber is also not acceptable owing to the fact that some degree of compression is required when load is applied. Bearing in these two conditions in mind, the above said rubber was chosen to be used for this device.

From this project, various rubbers will be considered to best suit the functionality of providing precise compression and hence an accurate reading of the force applied for the compression. Various factors are considered for selecting the material of the rubber. The density and the thickness of the rubber are the main factors which are considered as more important. The main types of rubber to be experimented are silicone rubber, PELITE rubber and natural rubber.

A displacement of 0.5 mm is the objective for the compression and that depends on the physical properties of the rubber. Some of the properties are defined and a comparison is made between different types of rubber. It is important to consider these values before selecting the rubber as they can make a major difference in the calibration and thus affecting the accuracy of the readings.

1) Hardness: It defines the resistance to indentation under specific conditions. The rubber is selected depending on the hardness value. Usually denoted in degrees on the Shore A^{o}. Lesser the degree, softer is the rubber. In this experiment, a medium rubber is been tested on. Soft rubbers can compress easily when minimum load is applied. If the rubber is too hard enough, then a much bigger load must be applied for a small deflection.

2) Load – deflection diagram: It shows the amount of rubber compressed when load is applied. So the graph basically depicts a deflection curve when load increases the deflection also increases. Depending on the physical properties of the deflection curve can vary in shape.

3) Load-deflection tolerances: The maximum resistance applied by the material when maximum load is applied to it.

4) Stiffness: A measure of the amount of deflection caused by the application of load. It is expressed in terms of Young's modulus. Slope of the load-deflection graph can indicates the stiffness of the material.

5) Compression: Many of the applications of rubber are for compression and it is necessary to define the maximum compression of the rubber for the loads applied for a specific point of time. The hardness and the strength of the material are primary factors that affect the compression.

6) Adhesion: Rubber to metal adhesion is crucial when designing devices with rubber. The rubber should be attached to the metal so that it does not get ripped off after long time usage.

7) Temperature responses. Rubbers undergo change in their physical properties when they are subjected to change in temperature. They usually become harder and stiffer with decreasing temperature, there by becoming less resilient. Rubber should also have operational stability in high temperatures

and should be resistant to higher temperature degradation. So the temperature range is important for the working of the transducer.

4.4 OPTOELECTRONICS

The whole concept of making a new force sensor has been to bring out a novel design using an existing technology. As stated in the previous chapters, the currently available sensors in the market are made with force sensitive resistors, capacitors and are fabricated using MEMS technology. But because of the shortcomings in these technologies, a new technique using optoelectronics has been investigated and devised to prove that optoelectronics can be implemented as a better solution for the existing ones.

Optoelectronics is the branch of physics that deals with the application of electronic devices used to control the source and detect the light. Light here refers to the broader spectrum of rays which includes x-rays, ultraviolet and infrared. Their application includes optical-to-electrical and electrical-to-optical purposes. They have a wide range of applications which ranges from light sensing to opto-coupling.

4.4.1 INFRA-RED EMITTING DIODES

The most commonly used optoelectronic devices when it comes to light emitting and sensing are LED's and phototransistors. LED'S are solid state P-N junction diodes that radiate light when it is forward biased. LED's are used mostly for illumination purposes and reflected light applications. Since LED's are small and efficient, they can be even incorporated into small size boxes. They have a longer lifetime and hence it is feasible to design them for a long time application devices. They are also compatible with solid state device circuitry

The PN junction is formed by doping one region of the semiconductor material with donor atoms and the adjacent region with acceptor atoms. When forward biased, the donor atoms attain enough energy to overcome the potential barrier. On crossing the barrier the carrier electrons will combine with the acceptor atoms and emit a photon of light.

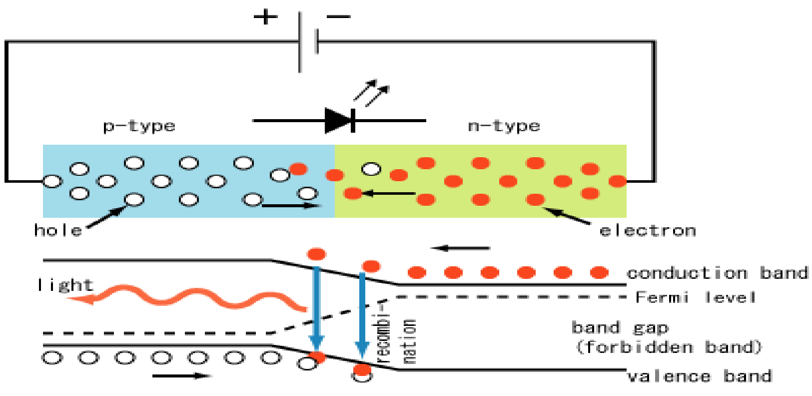

Figure 4.11 Working of IRED

The spectrum of light emitted can be varied by the semiconductor material and how it is processed. The intensity of light emitted is directly proportional to the current flowing to the diode. LED's emit rays in a narrow spectrum of wavelength 400nm - 700nm. Other than the visible light, LED's also emit light in the infrared region ranging from 750nm to 1mm. These are specially called as Infrared Emitting diodes. Since the wavelength of IRED's matches the peak spectral responses of the photo transistors, GaAs or GaAlAs IRED'S are used as emitters

4.4.2 <u>PHOTOTRANSISTORS</u>

Photo transistors are semiconductor devices that are in essence NPN transistors which are sensitive to light incident on base-collector base. Their radiation frequency response peaks are in the infrared region and to match their response, the emitters are always infrared emitting diodes. By doing this, it is possible to increase the signal-to-noise ratio. Phototransistors possess an internal gain and hence are better qualified in sensitivity when compared to other photodiodes. GaAlAs diodes are high gain, low noise photo transistors designed for better sensitivity.

Basically photo transistors can be any bipolar transistors that are encased in a transparent case so that light incident on the base-collector junction generates a base current. The actual working of the photo transistor depends on the biasing arrangement and the light intensity falling. There are two modes by which the phototransistor can be connected. When connecting the PN junction in forward bias, the current through the junctions due to the incident light will be irrelevant. However, when the phototransistor is connected in reverse biased, the current induced from the base increases and will be a function of the incident light. So the normal mode of operation is the reverse bias condition (How phototransistors operate., 2012).

If the collector base diode of the bipolar transistor is configured to be PN junction, the light induced current now, replaces the base current. As per the characteristics of a transistor, increase in base current can cause a corresponding change in collector current. Hence depending on the amount of light falling on the base, it can cause an increase in the base current which in turn cause increase in collector current (How phototransistors operate., 2012)

Few advantages of using photo transistor are;

1) Low cost and near IR detection.
2) High gain.
3) Usable with any source of light.
4) General characteristics match that of other signal transistors.

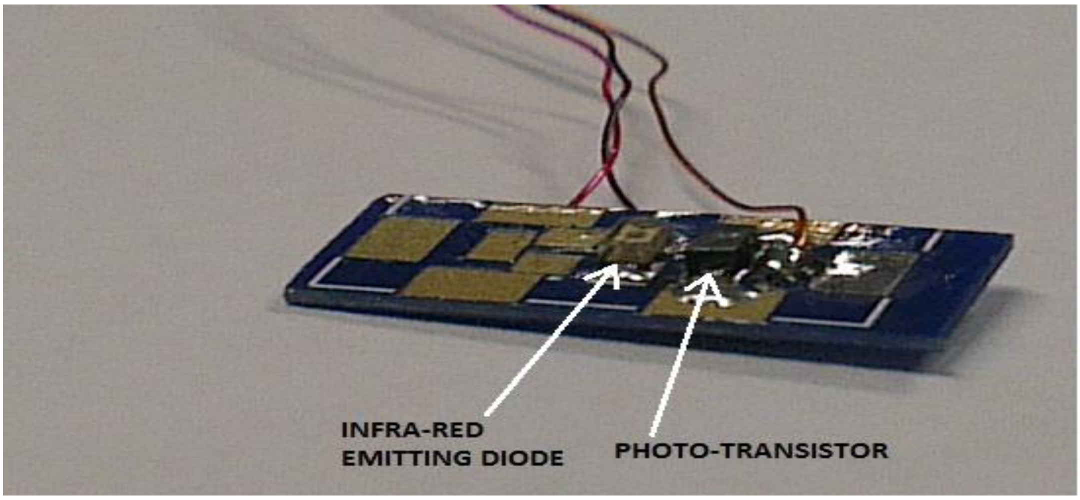

Figure 4.12 Electronic circuit setup

The size of the circuit board, seen in fig. 4.12, is 17 mm X 7 mm X 2mm. The height of the diodes is 1mm, which makes the total height of the circuitry to be 3mm.

5 METHODOLOGY

5.1 OVERVIEW

The in-sole method has been chosen over the force plate techniques since continuous steps in the gait can be recorded. An instrumented insole can be the best device to measure the foot-loading units while doing a normal activity. Since the size of the insole is been limited to very small thickness, it can slip into any kind of shoes without being uncomfortable to the patient. The pressure is varied from a low pressure to the maximum pressure.

5.2 OPERATION

5.2.1 ELECTRONIC CIRCUIT

An emitter-receiver combination is used here as an optical sensor. They are separated by a small wedge so that the receiver does not detect the light from the emitter. This can prevent the receiver from giving false readings. The emitter used here is VSMBN1940X01(Semiconductors, n.d.-a), a high-speed infrared emitting diode with wavelength 940 nm in GaAlAs double hetero technology with high radiant power molded in clear untinted plastic package. Features of this diode include high reliability, high radiant intensity, angle of half sensitivity, $\theta = \pm 60^0$, suitable for high pulse current operation. With these features, it can be used as a high transmissive sensor. Along with it, a silicon phototransistor TEMT7100X01 (Semiconductors, n.d.-b) is paired with it. This phototransistor can detect the infrared rays. It has high sensitivity and the filter bandwidth is matched with 830 nm to 950 nm IR emitters. The emitter-detector pai can be replaced by a single photomicrosensor EE-SY193. This sensor has both the emitter and detector incorporated into a single chip thus minimizing the utility of space. However, the height of the sensor is 0.95mm, which is slightly longer than the former sensor. Either of the two diodes can be used for the experiment.

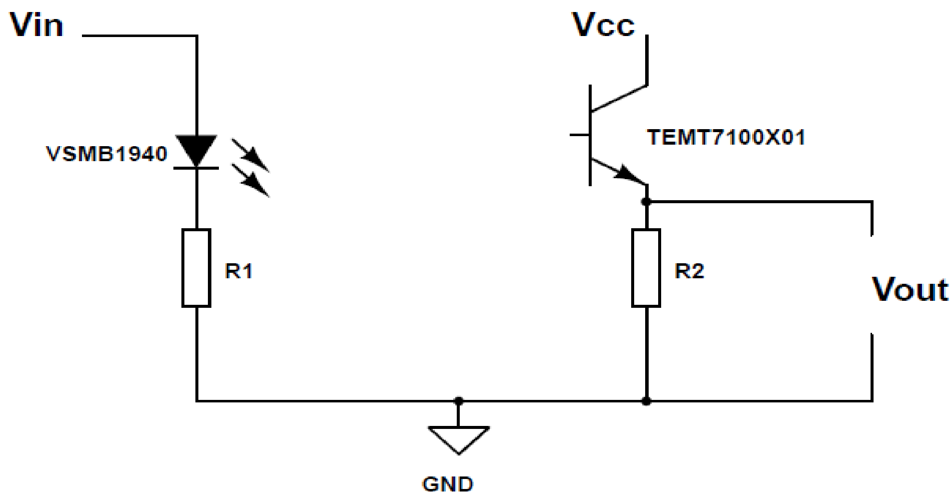

Figure 5.1 Circuit diagram

The optical distance sensor is placed within an closed shell, where the inferior surface of the in sole is coated with a reflective material allowing the IR light from the emitter be reflected from the reflective surface and gets detected by the receiver. In normal working, the detector is sensitive to a reflective angle of ± 60^{0}. Once the device is compressed, the angle of refraction changes and the intensity of light detected also changes. This change of angle will change the output voltage linearly. The analogue voltage can be converted into a digital voltage for further processing and then displaying it on the display monitor. The whole procedure can then be repeated with different pressure ranges. In this project, the digital output voltage will be observed in an oscilloscope or a digital voltmeter.

5.2.2 MECHANICAL

In order to demonstrate the application of optoelectronics to measure plantar pressure, a prototype device capable of fitting into the shoe has been developed. The device consists of a mechanical arrangement made of aluminium with a piece of rubber sandwiched between the two components of the device, as shown in fig.5.2.

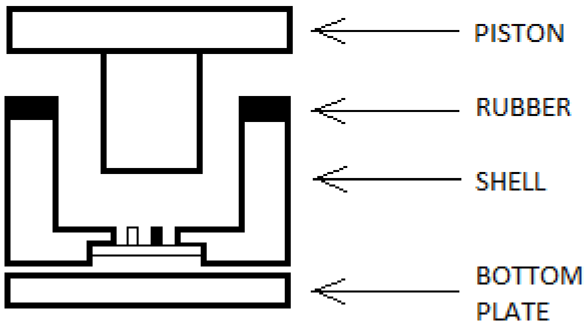

Figure 5.2 Mechanical setup

The device is 28.61 mm in height and 28mm in diameter. The thickness of the rubber is around 6.5 mm and is attached to the shell with two way stick tape. The electronics is placed into position through the slot and is held in position by the recess. The wires from the electronic circuit are connected to the power supply and output is observed either in an oscilloscope or in a digital voltmeter.

To know whether the theoretical design was responding to the loads, the device had to undergo three different compressive tests. Initial experimentation was carried out to prove that device actually worked when load was applied, by observing the output voltage. Static loads were applied to the sensor with the help of a loading instrument. Since the transducer was going to be experimented for the first time, load range from 0 N to a maximum of 300 N was decided to be used.

5.3 PROCEDURE

The circuit is connected as per shown in the fig. 5.1. For the initial calibration, the infrared emitting diode VSMB1940 is triggered by applying an input voltage. Once the diode gets triggered, it starts emitting infra-red (IR) rays. Thee rays gets deflected after colliding on the reflective surface of the piston. The rays are then detected by the photo transistor TEMT7100. When the IR rays falls on the base of the phototransistor, the subsequent change in voltage can be perceived from the oscilloscope. Initially the experiment is conducted in a no-load condition. Later an initial compression of 200 kPa is loaded onto the transducer. The change in the voltage is seen on the oscilloscope. Subsequently the load applied is increased gradually till 800 kPa. The corresponding voltage changes are plotted in a graph and the linearity of the system is checked. The experiment is repeated for different thickness of rubber and the linearity graph is plotted.

5.3.1 STATIC TESTING

In the initial test, calibrated load plates were applied manually over the sensor to obtain a relation between the applied load and output voltage of the circuit. The load plates used were man-made weights designated between 900 grams and 1 kilogram. The load is incremented every 1 kg up to 20 kgs and the respective voltage readings are recorded. The correlation between the applied load and the output voltage is obtained. The changing voltage can be seen in the oscilloscope as a waveform. The experiment is repeated by loading the transducer with weight up to 20 kilograms and then back to the initial condition. For each of the load, the voltage is observed and the graph is plotted. The experimental setup can be seen in fig 5.3.

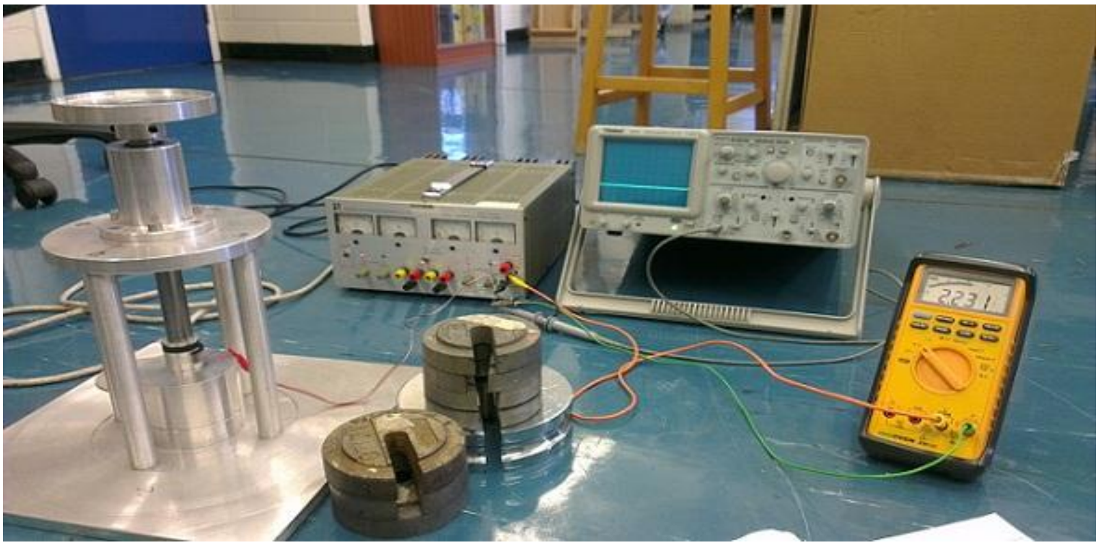

Figure 5.3 Experimental setup for compressive test 1.

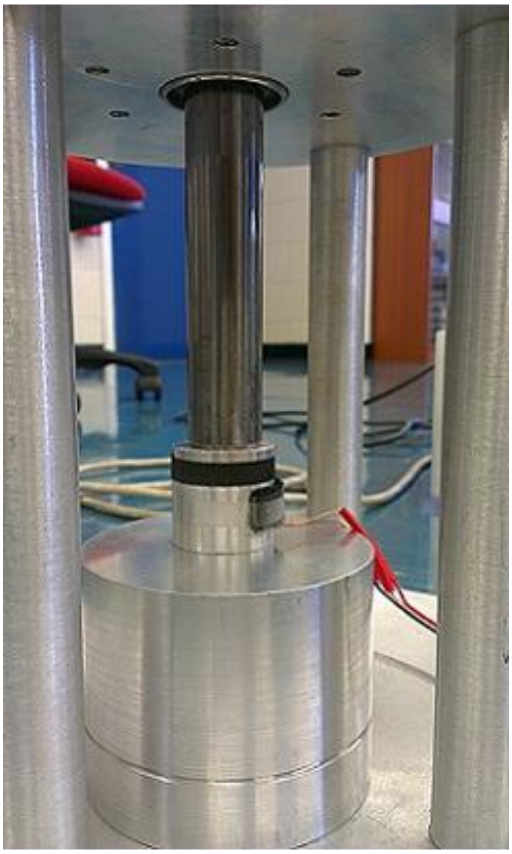

Figure 5.4 Loading of the transducer for compressive test 1

In fig. 5.4, the loading of the transducer is shown. As load plate is added on top of the piston, the piston pushes the rubber down, thus showing the compressive action.

The second compressive test is conducted with the help of INSTRON ELECTROPULS SYSTEM™ E10000. Since the range of load to apply is a maximum 300 N, a load cell of 1kN is used for compressing the transducer. A grip which has a flat base is fitted to the load cell so that there is an evenly pressure distribution on the transducer. The sampling frequency is set at 10N/sec and the sampling rate is 10 Hertz. The instrument is run with no load condition initially, and then incremented every 10N, while simultaneously the voltage reading is taken with the displacement recorded for further analysis. The experimental setup is as shown in fig. 5.5. Figure 5.6 shows how the load cell is loaded on the transducer. The base of the load cell rests on top of the transducer so that there is uniform loading given to the transducer (fig 5.7).

Figure 5.5 Loading setup for compressive test 2

The third compressive test is almost similar to that of the second one, only which this time the weight is increased up to 300N and readings taken simultaneously with pressure applied and incremented every 10N.

The final test is conducted to check whether the saturation of the output voltage around 300N that was obtained in the previous test was either due to the saturation of the detector or because of a mechanical snag. The probable mechanical snag can be explained as in fig. 5.8.

For each of the test, load versus the output voltage is plotted. For the saturation test, load versus displacement is plotted.

Figure 5.6 Load cell with the transducer

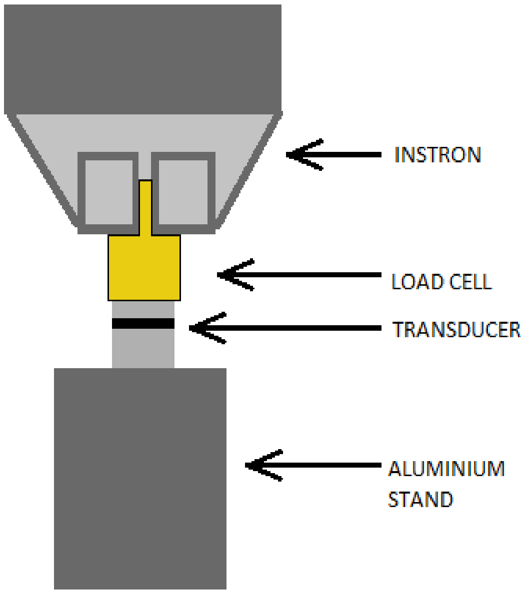

Figure 5.7 Load cell loaded on transducer

From the previous test, it could be observed that the output voltage was getting saturated around the 270 N – 300 N load ranges. To substantiate this interpretation, test were conducted by applying load onto the device. Force up to 300N was applied and the compression of the rubber was observed along with the downward displacement of the piston (fig 5.8). The readings were recorded and graphs have been plotted to support this work.

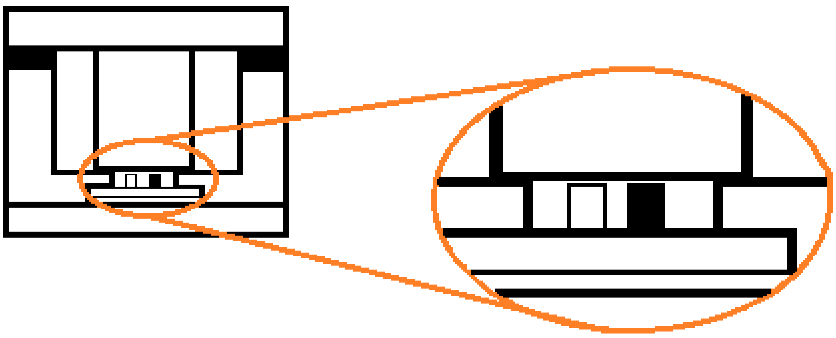

Figure 5.8 Piston resting on a recess a possible reason for saturation

6 RESULTS AND DISCUSSION

6.1 CALCULATIONS

From the experiments conducted, it can be seen that the transducer has responded to different loads ranging from 0N to 300N. The response of the sensor has been shown in voltage and displacement. The characteristics of the rubber can be assumed if the tensile strength of the rubber is calculated.

We know,

Pressure is calculated as force applied per unit area. The SI unit of pressure is pascal, (Pa).

So mathematically, it can be expressed as;

$$Pressure = \frac{Load}{Area} \quad (6.1)$$

The hardness of the rubber can be calculated, considering a load of 200N applied and from the displacement is obtained the graph. The piece of rubber is affixed on to the shell and takes in a ring structure. The outer and the inner diameter of the ring are measured to be 28mm and 20mm.

Therefore the

$$Area\ of\ the\ ring = \frac{\pi}{4} \times (outer\ diameter)^2 - (inner\ diameter)^2 \quad (6.2)$$

$$= \frac{\pi}{4}(28)^2 - (20)^2 \quad (6.3)$$

$$= 301.44\ X\ 10^{-6}\ mm^2. \quad (6.4)$$

Given, Load = 200 N

Hence,

$$Pressure = \frac{Load}{Area} \tag{6.5}$$

$$= \frac{200}{301.44 \times 10-^{6}} \tag{6.6}$$

$$= 663\ kPa \tag{6.7}$$

The main aim here is to find the shore hardness of the rubber. To begin with, we have,

$$Young'smodulus = \frac{Stress}{Strain} \tag{6.8}$$

Stress is equivalent to the pressure = 663 kPa.

General equation of strain is given by

$$Strain = \frac{Reduction\ in\ thickness}{Original\ thickness} \tag{6.9}$$

The thickness mentioned here is the thickness of the rubber. The original thickness of the rubber is found to be 6.5 mm and the reduction in thickness is found to be 0.77 mm after applying 200 N load,

Substituting the rubber thickness in equation (6.9);

$$= \frac{0.77}{6.5}\ mm \tag{6.10}$$

$$= 0.118\ \text{mm} = 0.118 \times 10^{-3}\ \text{m} \tag{6.11}$$

Since we have both the quantities, Young's modulus can be calculated from equation (6.8),

$$= \frac{663 \text{ kPa}}{0.118 \times 10^{-3} \text{ m}} \tag{6.12}$$

$$= 5.618 \text{ N mm}^{-2} = 5.6\ kPa \tag{6.13}$$

6.2 OBSERVATIONS

The series of graphs have been illustrated which gives a better representation of the systems functioning. The curves from the graphs can be analysed to understand how the system responds to different loads. The graph has been plotted with load (kilograms) in the vertical axis against voltage output (V) in the horizontal axis. Graphs showing the displacement, in mm of the rubber with the compressed load, are also given.

6.2.1 COMPRESSIVE TEST 1

As seen in the graphs, it can be noticed that, the loading stage takes the lower part of the curve while the unloading stage follows the upper side of the curve.

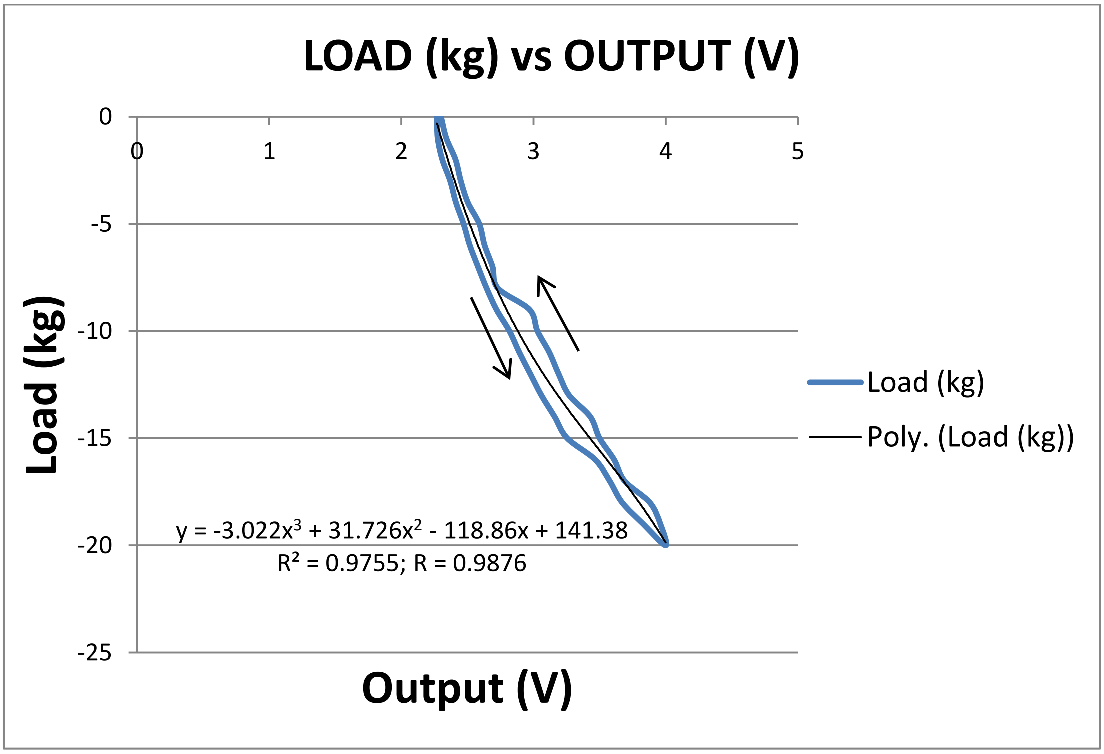

Figure 6.1 Voltage response to load for compressive test 1

Figure 6.1 was obtained after plotting load and output voltage. A third degree polynomial regression was fitted to suit the curve obtained. The test was conducted by loading the sensor with calibrated weights and observing the change in output voltage. The graph shows the voltage response when loads were added up to 20kgs and then back to 0kg. The correlation efficient R = 0.9876, is almost near to one indicating that the system line is almost following the points in the graph. In no-load condition, the voltmeter showed an initial reading of 2.224V. As the load increased, the output voltage also increased and reached up to 3.960 V for 20 kg, where this is represented by the downward indication of the arrow. While unloading, the curve follows a secondary path and is indicated by the ascending arrow.

After the initial testing, the system was given few minutes break, so as to allow the rubber recuperate from the initial loading compression. The experiment was repeated using the same calibrated loads within the same time frame. To best fit in the curve, a third degree polynomial was used. In this it can be observed that the R coefficient (R = .9905) had come closer to 1 showing the polynomial fits better this time and closely represents the obtained curve. For zero-load, the voltmeter showed a reading of 2.260V and finally for 20kgs, the output voltage was found to be 4.000 V. The compression curve, fig 6.2, has been represented by the direction of the arrows.

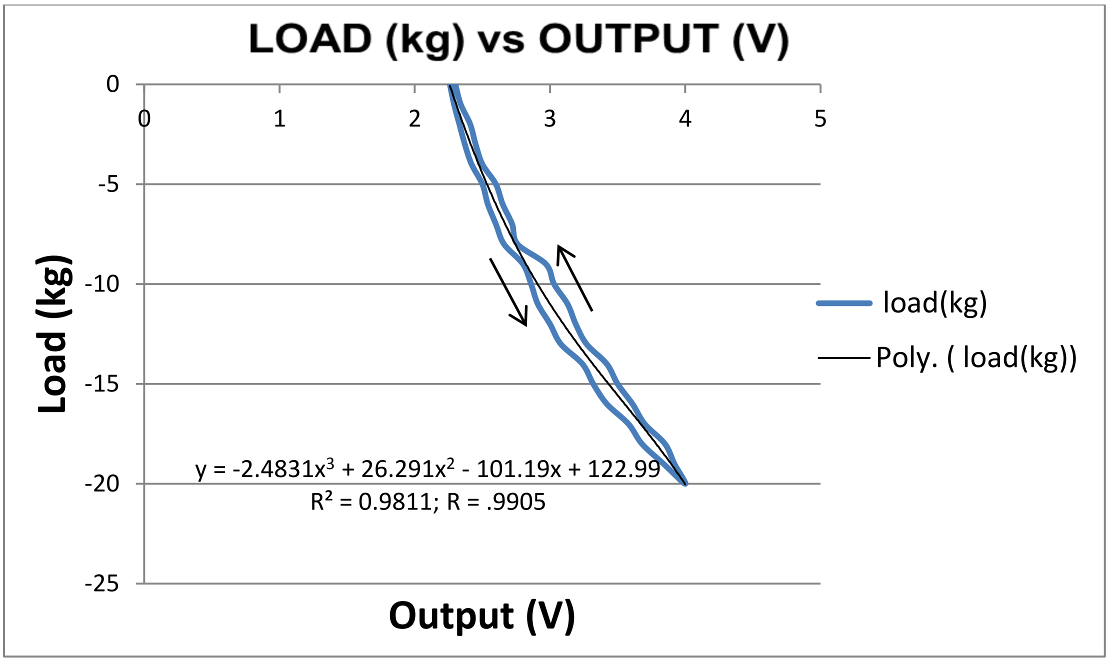

Figure 6.2 Voltage response to load for compressive test 1 after few minutes break.

6.2.2 <u>COMPRESSIVE TEST 2</u>

Once it was confirmed that the system was responding to loads, the tests were conducted in a much more specific and precise load controlled conditions. INSTRON ELECTROPULS E10000 was chosen as it had the following options;

1) To determine the intensity of load to be applied,
2) The duration of the weight to be loaded on the transducer,
3) Time window between every load increment and
4) Time period for running the entire test.

The following graphs were obtained from the INSTRON test. The transducer was loaded in a controlled method and the output readings were simultaneously recorded as the load was applied on the transducer.

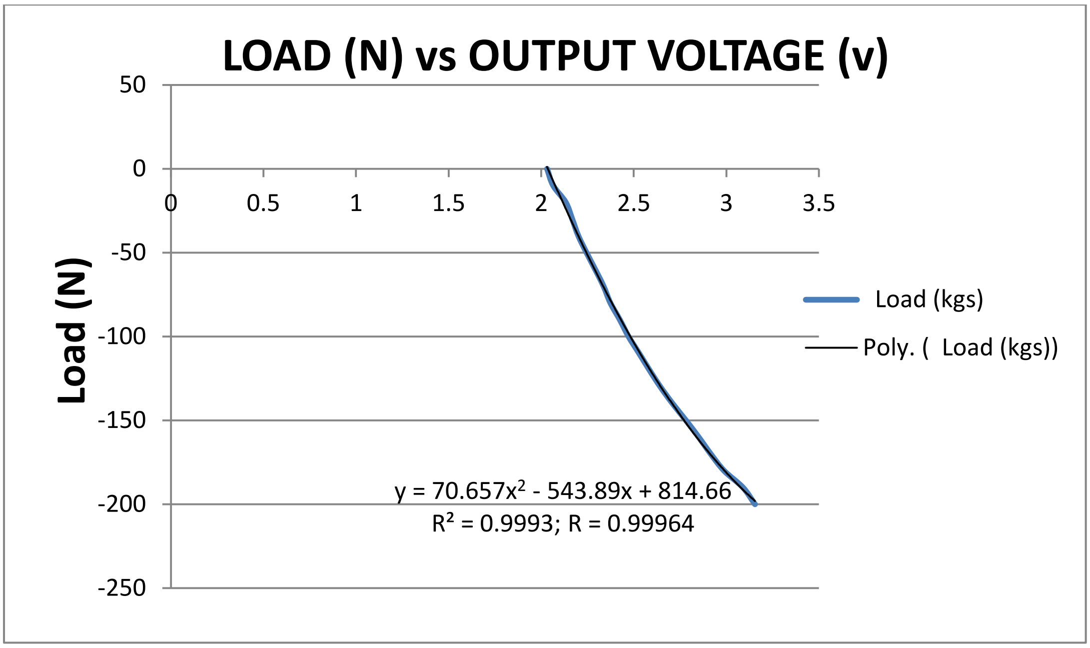

Figure 6.3 Voltage response to load for compressive test 2

Here the load was applied to 200 N and then returned to the initial no load condition within 30 seconds. Figure 6.3 shows the load (in Newton) plotted against output voltage (V). The initial voltage was 2.032V and by 200N it was around 3.155N. A second degree polynomial is applied to the curve to best fit the response. Since the load is controlled and precise, it showed a non-linear relation between the applied load and output and this is indicated by R = 0.99964 implying the data points did not fit into the regression line.

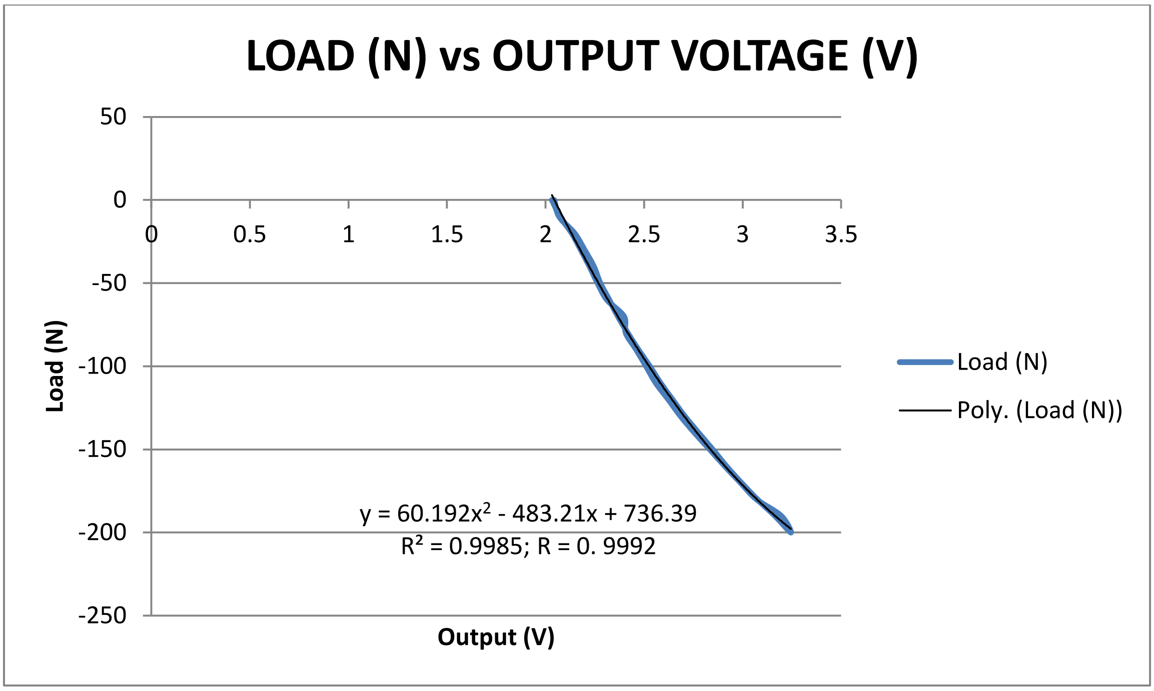

Figure 6.4 Voltage response to load for compressive test 2 loading up to 200N

Figure 6.4 shows the voltage response when the transducer was compressed to 200N.in a series of steps of 10N from 0N with a frequency of 10N/second. The following graph is obtained after plotting load and output voltage. A second degree polynomial regression was fitted to suit the curve obtained. Since the correlation efficient R = 0.9992, was approximately near to 1 proved that the data values were almost fitting the polynomial line. At 0N, the voltmeter showed an initial reading of 2.032V. As the load increased, the output voltage also increased non-linearly and reached up to 3.245 V for 200N. Since the system was linear the hysteresis curve was absent.

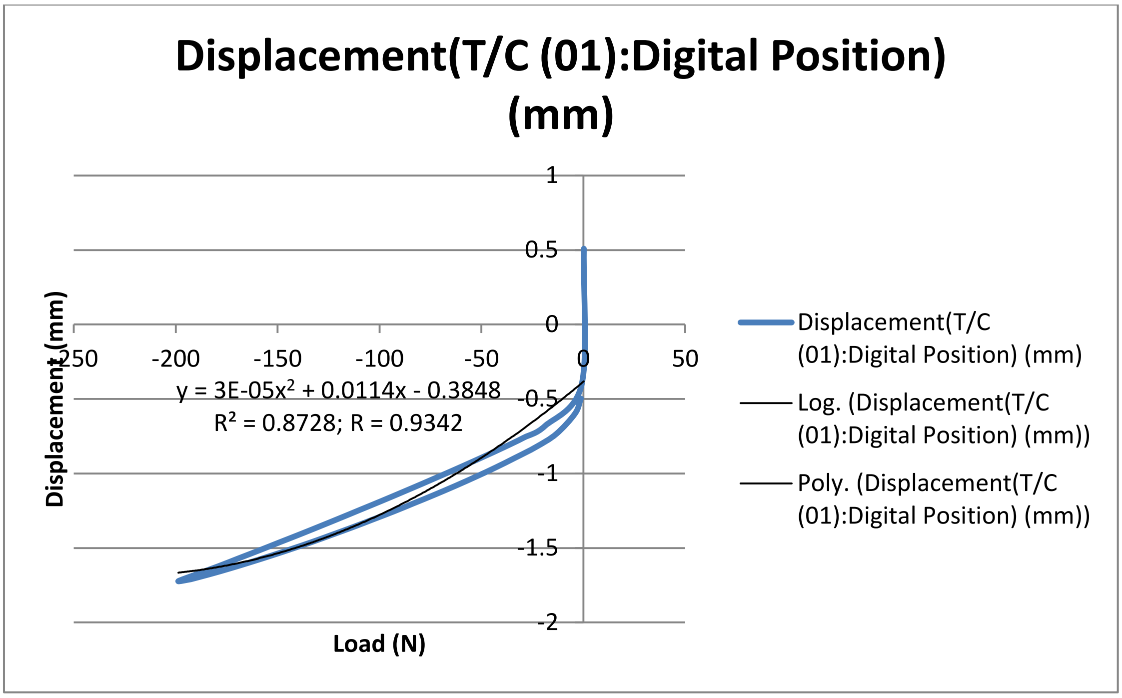

Figure 6.5 Displacement response to load for compressive test 2

The displacement of the rubber has been recorded and plotted in the succeeding graph (fig. 6.5). The gradual decrease in thickness of the rubber can be observed as it gets compressed. Force is applied to 200 N and the rubber gets compressed to 0.77mm of the total gap of 2mm. The curve has been fitted with a second degree polynomial and the R value (0.9342) shows that the displacement has a hysteresis curve.

6.2.3 <u>COMPRESSIVE TEST 3</u>

The succeeding graph shown in fig. 6.6 was obtained when the voltage response was plotted for 0 – 300N. A second degree polynomial regression was fitted to suit the curve obtained. The graph shows the voltage response when loads were put up to 300 N and then back to 0N. The correlation efficient R = 0.9987 is relatively near to 1, indicating that the line was fitting the points obtained from the experiment. In no-load condition, the voltmeter showed an initial reading of 2.032V. As the load increased, the output voltage also increased and reached up to 3.814 V for 300 N. But a main feature was the presence of a saturation curve which was observed after 270 N, where the output voltage remained to be constant at 3.814V.

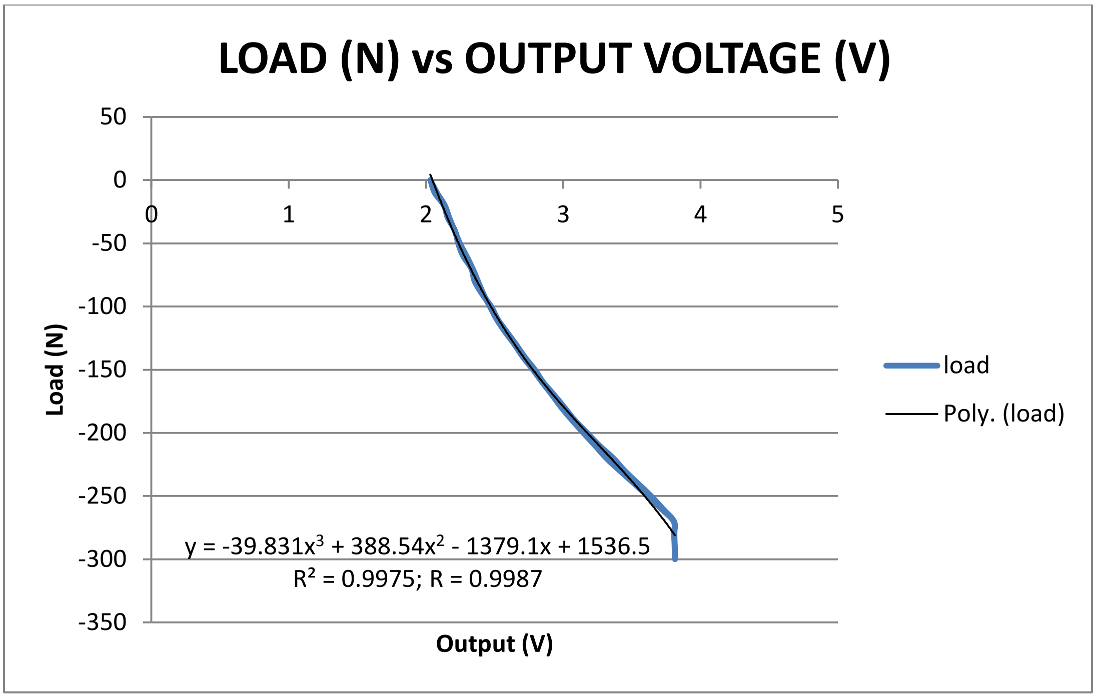

Figure 6.6 Voltage response to load for compressive test 2

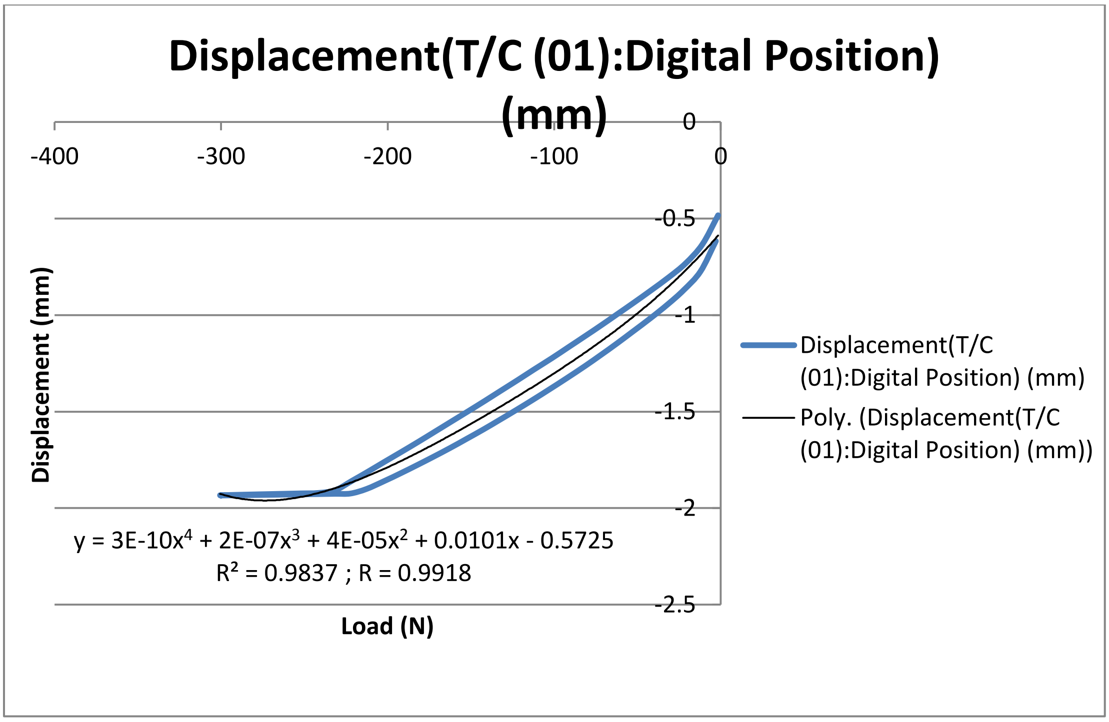

Figure 6.7 Displacement response to load for compressive test 2

From the above graph, since it was observed that the output voltage was getting saturated after 270 N, the displacement test was conducted to check whether the rubber was getting saturated or not. Graph, fig. 6.7, is the displacement curve fitted in with fourth degree polynomial trend with R = 0.9918. It can be stated that at 270 N, the displacement becomes a constant at 1.93mm. The R value being low also implies that the displacement curve is not linear and that the rubber when decompressed does not follow the same path as when it got compressed.

These graphs show a rough indication of how the system works and the response of the voltage to the force applied. The displacement curve has also been plotted to indicate characteristics of the rubber to different pressure loads. Primary results show that the compression of the rubber is not linear. Further arguments about the rubber have been discussed later.

6.3 DISCUSSION

Designing a pressure transducer using optoelectronics for plantar pressure measurement has been a challenging task. Although theoretically the concept was realistic, the initial stage of experiments proved that it is practically achievable also. Starting from the basic design to the manufacturing stage, numerous factors had to be considered which had to follow the fundamentals to fit the criteria of being a pressure transducer. The main points to be considered were the size of the transducer, the ability of the transducer to measure only vertical forces, while overlooking the other degrees of motion, range of load measured. A small prototype design was primarily considered. Even though there were many mechanical examples which provided only vertical motion, they were impractical to implement as the tiny mechanical structures would be highly unstable under heavy loads. Hence it was decided to use aluminium as the main frame. Use of rubber has been highly investigated here since the transducer works on the compression of the rubber. The rubber used here is an industrial rubber of Young’s modulus of 5.6 kPa. This fairly shows that the rubber is stiff and hard. Even though it was a trial, the rubber showed impressive characteristics. The rubber so chosen was not soft and not hard enough.

The basic working principle of the pressure transducer has been aimed on the electronic circuitry. An IR emitting LED VSMB1940 is used with a matching photo transistor TEMT7100. These diodes are surface mount devices and work in 940nm range. The IRED used here has high radiant energy, transmissive power and works well in low space applications. Phototransistor TEMT7100 pairs well with VSMB1940 since the daylight blocking filter matches with the TEMT7100. The use of optoelectronics has been a novel thought and the purpose of application of the model has come out well. Though the output voltage from the photo detector was not linear to the applied force, it can be designed to be more sensitive to the compressed load. This can be done by changing the resistor values to adjust the sensitivity of the circuit.

The graphs showed an overall concept about the working of the pressure transducer. It plotted the response of the photo-detector when force was applied, measured in voltage. Figure 6.1 and 6.2 shows the loading response of the transducer with output voltage. Load was applied from 0kg to 20 kg, back to the minimum load and the hysteresis curve was obtained. The probable explanation for the presence of two curves could be because of dissimilarity in the denominations of the load plates, for example 1kg load showed a weight of 900g and so a load of 2kg was found to be 1800g. Hence the loads applied were found to be non-uniform and this could have bought erroneous results. Secondly, it was impractical to get the exact readings manually as pressure applied to the transducer was continuous and could not be controlled. So the output voltage kept on varying, unable to take specific output readings. From both the graphs, the system tends to be linear at lower loads, but as the load increases, the system tends to drift away from the linear line.

To be more specific about the test and have better clarity in the output, the tests are done under load controlled conditions. The INSTRON is chosen for the fact that the intensity or load applied can be controlled with the help of the software. Unlike the load plates, the INSTRON is able to deliver precise loads and hence the readings would be accurate.

Figures 6.3 – 6.7 represents the analysis of the test performed using the INSTRON. Initial glance on the graphs proves the fact that instrument testing is more accurate and specific when compared to the load plate method. The graphs become more linear and the hysteresis curve is almost absent. The transducer responded well in the range 0 – 200 N and better till 270 N, thereafter which the transducer output starts getting saturated. The plausible reasons for the saturation effect might be the fact that either there was no further compression of the rubber on increasing loads or the photo-detector could not detect the reflected IR rays. This effect could be seen from the figure 6.7 (load-displacement graph of 300N) which shows displacement getting constant at 1.93 mm.

7 DESIGN AND EXPERIMENTAL ISSUES

Before the final design was confirmed, the design for the transducer had to undergo a number of changes. The design for the mechanical and the electronics setup was been reviewed time and over again. A number of various designs were considered which had to satisfy the condition of uniaxial-motion in *y* direction. Several design concepts were bought in to accommodate the conditions required.

7.1 DESIGN

The mechanical design had to be reviewed by the experts for finding out the faults and also to understand the feasibility of the design. Few drawings were initially designed for the sensor. Different concepts were discussed which included a linear gear and pinion system, a linear linkage design et cetera. Each of the design had faults and short comings, which had to be solved and include better changes that made the design much more feasible.

To start with, a design with the linear linkage was the first sketch created. However the sizes of the linkages were the main concern. The overall size of the transducer had to be small enough to fit under the sole of the shoe and the size can be stated in millimetres. So when the intensity of the load to be applied was taken into consideration, it was understood that the linkages would not be able to withstand high forces. There were maximum chances for the linkages to break and collapse totally under heavy weight. Hence further discussion of the design had to be put off.

To reduce the number of linkages in the system, a secondary design was drafted using the gear system. A vertical gear system referred as the *rack and pinion* was proposed. Advantage of using a pinion gear system for this device would be that it requires only a smaller gear of pair, which is acceptable in terms of size (“Springs and gears,” n.d.). Even though the gear system is compact and allow short distance movements, they need to be precisely cut using special machinery (“Springs and gears,” n.d.). Just as the previous one, the mentioned design too was fragile enough to take in the loads. Also, it was a challenging task to manufacture the small gear system using the special machinery which can increase the initial building cost. Since a uniform vertical motion was expected, the gear system had to be supported by additional bars on four sides, which made the design more complicated.

Henceforth it was decided to exclude design that had micro structures for a transducer design and to make it as simple as it can. Later a square *box-within-a-box* design was drawn, which had a technique to restrict 5-degrees of freedom and allow only vertical motion. This design was almost similar to the current design. But for the ease of manufacturing and perfect positioning under the contours of the foot, circular design was accepted and considered for manufacturing.

Once the design was fixed, a suitable material had to be chosen for the casing of the transducer. Two materials were considered for this objective – Acetal and Aluminium.

Acetal is engineering thermoplastic. It has excellent mechanical properties which makes it a very good material for constructing the device. Acetal has good tensile strength and stiffness and a very good dimensional stability. Since it is a tough material, it can protect the objects it is required to safeguard. Moreover it is easy to manufacture and process and low friction and wear properties (http://www.acetal-bar.co.uk/, n.d.). However acetal is a translucent material and the infra-red rays can pass through it making the photo detector was sensitive to the ambient light, which could affect the accuracy of the system. Hence aluminium was considered as a better material to manufacture the device.

7.2 RUBBER

The main component aiding in the vertical motion is the rubber. The rubber gets compressed when load is applied on the piston and this is the key mechanism for the overall working of the arrangement. As stated in the previous chapter, the rubber had to specifically be chosen with the main focus kept on the properties of the rubber. Primary condition was the compression factor and secondary was the thickness. Thickness had to be made minimal and the compression of the rubber had to be in an intermediate level. The rubber could not be soft enough where the piston would touch the electronics and could not be hard enough where rarely any motion was recorded. Hence different rubber of different material and thickness had to be tested and evaluated. In connection with this, the length of the piston had to be adjusted so that a minimum distance between the bottom surface of the piston and the top surface of the diodes was kept constantly at 2mm when no load was applied.

7.3 ELECTRONICS

Since the overall system was small in size, the dimensions of the circuit board were also an important factor. The connection wires providing the input voltage and acquiring output were of 1mm in diameter and that took space while designing the internal structures of the shell. Later on these wires were replaced with strain gauge wires which were 0.1mm in diameter.

The connection wires soldered to the electronics were big in diameter when compared to the overall dimensions of the circuit board. This was a problem when considering the design of the shell. Later the thick wires were replaced with strain gauge wires which were few millimetres in diameter.

The wires that connected the electronic circuit to the power supply and the oscilloscope were thin and they were susceptible to bending and breaking. Therefore a slot, for taking the wires, in the shell was considered during the designing time. The slot given for the wires is now bigger in dimensions when compared to the size of the wires, so provisions have to be made to cover the slot so that no light can enter and alter the readings.

8 CONCLUSION

8.1 INFERENCE

Plantar pressure measurements are well recognised and a useful method to understand the human gait system. Although there are lot of pressure measurement techniques, it solely depends on the method of choice and details of analysis recommended by the physician. There have been many techniques experimented in the past, while some of them have been used in the present market. Pressure platforms have given the qualitative side of the plantar pressure, but most of the in sole systems have been successful in proving the quantitative measures of foot pressure. The design of a new in-sole transducer using an optical distance sensor should be an innovative one and should be aspiring enough to take pressure measurement to the next level. Further experiments are yet to be done to substantiate the practical application of this novel transducer.

8.2 FUTURE WORKS

Since this is the initial testing of a novel transducer, a lot of advancements and changes can be applied into the further developments of the transducer. First and foremost would be to decrease the size of the transducer and make it as small as possible. Most of the present day pressure transducers available are small in size so that it can fit either into the in-sole or in the socket. Similarly, the main aim or the challenge for this project would be to reduce the dimension and make it practically miniaturized. This should help in using the transducer for socket interface pressure measurement also. With decreasing size of the transducer come associated changes. The size of the circuit board and thickness of the rubber will also decrease, thereby decreasing the overall size of the transducer. Smaller diodes got to be used that have smaller footprints. (Footprints refer to the dimension of diodes). The current ones used are 0805 footprint diodes. So the next objective of this project would be to find smaller diodes that can be found, without sacrificing the intensity and power of the diodes. As the size of the rubber decreases, the mechanical properties of the rubber will also be affected in a limited scale. The thickness of the rubber will be brought down. So either the same rubber or a different rubber with better hardness will have to be experimented, so that the linearity of the system does not change. For the initial testing, loads up to 300N were tested. Next time, it should be such that the transducer is tested with much more heavier load, maybe up to 1000N, since the weight applied by a normal man falls in this range. The rate of sampling frequency will be increased to 50-100 Hz and more data points should be obtained which will help in plotting a better curve for analysing the working of the transducer with better precision and accuracy. Data interpretation has been a major tool in most of the biomechanical analysis and helps the clinicians to better understand the intensity of pressure applied by observing the pressure mapping. A similar technique would be attempted to merge data analysis with the transducer so that it gives in a real time examination for the clinicians to study the image.

The initial study has proved the theoretical concept and conducting further experiments with the changes mentioned in the preceding paragraph can substantiate the fact that the design can be made feasible and can be used as uniaxial force transducer and also for other arrangements, such as employing this with the socket of an amputee.

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