A Standard Wearable System for checking Gait Switching and Balance in Paraplegic Exoskeleton

Preethika Immaculate Britto[1],
[1]Assistant Professor,
Biomedical Engineering Division,
School of Biosciences and Technology, VIT University,
Vellore,

Vidhya.S[2],
2Assistant Professor Senior,
Biomedical Engineering Division,
School of Biosciences and Technology, VIT University,
Vellore,

Kalluri Yogeshwar Reddy3*
[3]B. Tech,
School of Electronics and Communication Engineering,
VIT University, Vellore, E-mail:

***Abstract*— This paper describes a wearable system capable of measuring many parameters relevant to gait analysis, and developed to provide quantitative analysis of gait outside of the confines of the traditional motion laboratory.It describes the development of a system for determining the pressure exerted by the exoskeleton platform by using a matrix of force sensitive resistors (FSR) placed under the foot plate and to get the inclination of ankle joint by using Inertial measurement unit (IMU) with respect to the horizontal standing position which will be used as inputs for a low cost lower limb exoskeleton for maintaining stability and to help the patient walk. The system is specifically for paraplegic patients based on the Austin project. It is designed for patients affected by paraplegia induced by spinal cord injuries at the level of the ninth thoracic vertebra and below. Such patients have good trunk, abdominal muscle and hand control but inadequate hip flexor and leg control making them unable to walk.**

Keywords—Exoskeleton, Paraplegia, Austin project, Wearable System, Gait Analysis

I. INTRODUCTION

Paraplegia often leads to complete lower limb paralysis and also possibly the trunk severely inhibits the mobility, independence and productivity of its patients. It is usually the result of spinal cord injuries or congenital conditions like spina bifida which affect the injury neural elements of the spinal canal. Spinal cord injury leads to paraplegia. It is important to note here that not all spinal cord injuries lead to paraplegia, the damage to the cord must be at the level of the first thoracic vertebra and below. At T1-T8 shown in *Fig .1* in some cases there is control of the hands, but poor trunk control as the result of lack of abdominal muscle control. Lower thoracic injuries (T-9 to T-12) allow good trunk and abdominal muscle control. Lumbar and sacral injuries yield decreasing control of the hip flexors and legs.

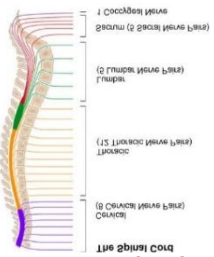
Fig.1 Spinal cord

Clinical research has identified clear links between human gait characteristics and different mechanical conditions. So exoskeletons are one way which provides mobility.While there are still many challenges associated with exoskeleton development that has yet to be perfected, the advancesin this field has been enormous.

There are two main approaches to exoskeleton development; there are technologies to augment the abilities of able bodied humans, often for military purposes and there are assistive technologies for physically challenged persons. Here, an exoskeleton can be defined as an active mechanical device worn by an operator fitting closely to his or her body that works in collaboration with the operator’s movements. The Austin project represents a series of technologies that have a deliberately stripped down clever design leading to low cost and therefore accessible exoskeleton systems for individuals with mobility disorders.

The fact that we as humans are bipeds and locomote over the ground with one foot in contact (walking), no feet in contact (running), or both feet in contact (standing) creates a major challenge to our balance control system. Because two-thirds of our body mass is located two-thirds of body height above the ground humans have an inherently unstable system unless a control system is continuously acting. This problem is even more pronounced in paraplegic patients as they lack adequate hip flexor and limb control. It is therefore of vital importance that human balance and posture control are studied if to design an effective system to improve the mobility of paraplegics.

Three major sensory systems are involved in balance and posture. Vision is the primarysystem involved in planning human locomotion and in avoiding obstacles along the way. The vestibular system is our 'gyro', which senses linear and angular accelerations. The somatosensory system is a multitude of sensors that sense the position and velocity of all body segments, their contact (impact) with external objects (including the ground), and the orientation of gravity.

II. HARDWARE/SENSORS

A. *Arduino*

Arduino can sense the environment by receiving input from a variety of sensors and can interact with its surroundings by controlling lights, motors, and other actuators. Ithas been used to process data collected from the IMU sensors as well as to run the necessary algorithms based on the data received to determine the motion tracking.

B. *IMU*

The IMU has been used for measuring the motion tracking response of the paraplegic exoskeleton for gait switching.

1. *Accelerometer*

An accelerometer is used to get proper acceleration. The acceleration measured by it need not be the coordinate acceleration (rate of change of velocity). For example, when no forces are applied on an accelerometer which is placed on the earth's surface will measure an acceleration $g=9.81\ m/s^2$ straight upwards, due to its weight. Force can be calculated by using basic formula of $F=ma$, where m is the weight of runner and a is the acceleration that has been obtained from the accelerometerconnected to the platform at the ankle.

2. *Gyroscope*

A gyroscope is a device for measuring or maintaining orientation, based on the principles of angular momentum. It has been used to measure the angles at the ankle during the gait.

C. *Force sensitive resistor (FSR) sensor*

The Force sensitive resistor sensor is used to detect the applied force. The force sensitive resistor sensor has been used for measuring the pressure exerted by the paraplegic exoskeleton for balance checking during gait cycle.

D. *Zig Bee*

Here, CC2500 Zig Bee module has been preferred which is a Single Chip, Low Cost, Low Power RF Transceiver and has an operating frequency range between 2400 – 2483.5 MHz. It uses FSK modulation technique for data transmission. It gives a 30 meters range with onboard antenna. It can be used on wireless security system or specific remote-control function and others wireless systems. This module supports various modulation formats and has a configurable data rate up to 500 K baud.

E. *Software*

ArduinoIntegrated Development Environment (IDE) is theprogram platform used to program the Arduino Uno board. The programming language is embedded C. Also MATLAB has been used to obtain the graphical representation of data.

III. IMPLEMENTATION METHODOLOGY

Fig. 2 shows the overall system block diagram. It consists of two units: First unit is related to data acquisition and the Second unit consists of the master controller part. Data acquisition part consists of the multiple sensor system (Inertial Measurement Unit, Force sensitive resistor sensor). In the second unit Digital data has been processed by Arduino and this processed data is transmitted through the Zig Bee wireless protocol. In the receiver section this data is received by another Zig Bee wireless protocol and then monitored on a personal computer for data analyzing.

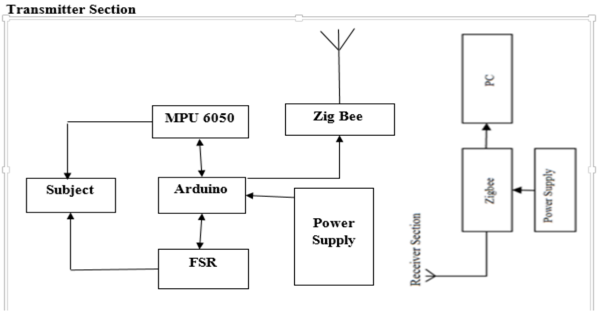

Fig. 2 Overall functional block diagram

A. *Data Acquisition Part*

There are fifteen areas on the sole foot shown in *Fig.3* that support body weight during normal gait. The heel and metatarsal and heel areas are selected to see if a person favors heel strike or forefeet strike. Adding two more area like mid feet and hallux would provide better results.The IMU was secured to the ankle joint.

Fig. 3: Foot pressure points

1. FSR Testing

Fig. 4 shows the FSR circuit with Arduino. The output voltage of pressure and positional coordinates of the FSR are obtained from seven positions on the foot platform. Four FSR's have been placed under the toe and metatarsal region and another two are placed close together under the heel.

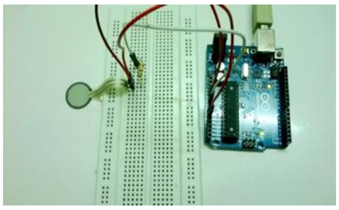

Fig. 4 Single FSR Circuit

To estimate pressure range two different caseshave been considered. Pressure is mass per unit area (p=w/A) so it bring to the conclusion that the max pressure is when a high mass is applied in a very small area. This is the case is when a very heavy person with weight of 300lb lands on his or her heel which is a very small are something around (1.98in x 1.98in), so it gives us maximum pressure of 76.5 psi. The minimum pressure is when a very light person with weight of 50lb and a very large plantar surface of (1.9in x 9.84in) lands on the floor with entire foot which gives a minimum pressure of 2.56 psi. So the ideal pressure range will be in the range shown in Table 1. *Fig. 5* represents the composite graphs of FSR calibration.

Table. 1 Pressure Range for FSR

Max pressure	Min pressure
76.5 psi	2.56 psi

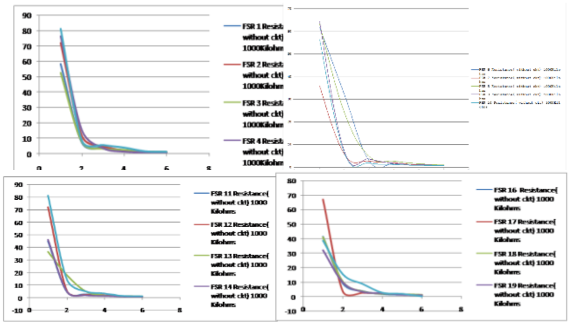

Fig. 5 Composite Graphs of FSR Calibration

Other than pressure range, our pressures sensors have to have:

- Accuracy within (5%)
- Corrosion and dust resistance
- High sensitivity
- Extremely thin (flexible and easy to adjust in shoe)

*Fig.6*shows the foot platform made of cardboard with the FSR matrix for pressure detection

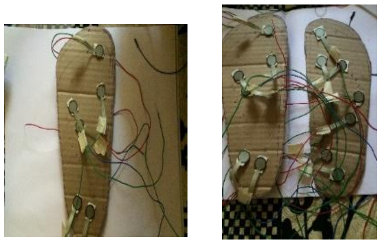

Fig. 6 Foot platform with FSR matrix for pressure detection

2. IMU Testing

The IMU circuit diagram has been shown in *Fig. 7*.
The IMU was secured to the ankle jointand the inclinations of the joint with respect to the horizontal standing position were obtained.

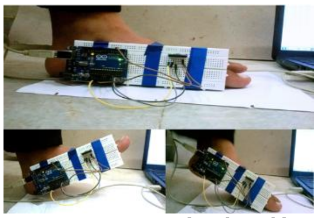

Fig. 7 IMU mounted to the ankle

For IMU, to be able to eliminate the noise**,**drift and get the accelerometer does notchange angle to detect other force than gravity**,** afilterhas been used called **'**Complementary filter'.
After application of filer and removing the noises, the IMU will start giving error free values. Although these values will contain some zero error or positioning error, to correct that we have chosen ground level as reference. At ground level, the inclination with respect to X Y and Z axis will be 0 degrees. At the initial testing of the IMU, when it is kept at ground the values were:X-14 Y-26 Z-56. So these will be subtracted from every reading in order to get 0 degrees at ground level.

IV. DISCUSSION

On the basis of results, firstly the foot platform to detect pressure were found to be functioning properly as the output voltage obtained from them is in corroboration with the theoretically calculated expected values. The output voltages obtained from the six FSR matrixes in quiet standing phase all have values above 4.5 volts. Since, the input voltage to each FSR is five volts, this makes sense. Also the resistance of an FSR decreases with magnitude of force applied on it, so when a man weighing seventy kilograms stands on a foot platform having six FSRs the resistances of the sensors go close to zero and the output voltage is supposed to be nearly five volts which is in agreement with our findings. Similarly in the toe off phase weights are detected only by the toe sensors T1 and metatarsal sensors T2-T4while the heel sensors H1, H2 detect no pressure at all. The output voltages obtained from T1-T4 not only give output voltages but these values are roughly above the values obtained in quiet standing phase since more weight is concentrated on fewer sensors. In the heel strike phase the weight of the subject is concentrated

ISSN: 2278-0181
TITCON-2015 Conference Proceedings

on the heel sensors H1, H2 and the toe sensors T1-T4 are not subjected to any weight. Consequently, we get output voltages in the range of 1890 to 4870 mV from the heel sensors while the toe sensors give zero output.

The three dimensional outputs were again obtained in the quiet standing, the toe off (ankle planter flexion) and the heel strike (ankle dorsiflexion) phase.During heel strike the ankle dorsiflexes and range of angle made by the ankle during dorsiflexion is 10-30 degrees. Similarly during initial toe off the ankle plantar flexes the range of angle made by the ankle during plantar flexion is 10 -15 degrees. The three dimensional coordinates obtained from the IMU in the quiet standing phase are all nearly zero which validates our result since the IMU was placed in its horizontal standing position parallel to the ground. In the toe off position theX and Y coordinates still remain close to zero but the Z coordinate is roughly -11 since the ankle is inclined downwards at an acute angle to the ground. Similarly in the heel strike position the ankle is inclined upwards justifying the values we get from the IMU in this case which are approximately (0, 0, 17).

V. FUTURE WORK

The design and fabrication of a system to measure the voltage output of foot pressure and to measure the inclination of joints in the leg to be implemented in a lower limb exoskeleton to help patients with spinal cord injury induced paraplegia at the level of T-9 and below is presented. The system will provide the magnitude of pressure at different points under the foot based on which the zero moment point is calculated. Whenever the ZMP shifts from the geometrical center of the platform compensation will be provided by the servomotors used in the exoskeleton to prevent the patient from toppling. It also gives the inclination of the joints. It is secured so that stability of the patient is ensured so as to prevent the subject from falling over when using the exoskeleton. This phenomenon will be used to help the exoskeleton be effective for climbing and descending gait. It is designed to provide the inputs to an exoskeleton which will use these values to perform its function of providing mobility to the patient. The system needs to be incorporated in the exoskeleton and tested on actual paraplegic patients to assess its performance and the necessary improvements will have to be made based on comparisons with the existing exoskeletons in the market and feedback obtained from the patients.

VI. REFERENCES

(1) Joint Torque and Energy Patterns in Normal Gait D. A. Winter and D. G. E. Robertson, Department of Kinesiology, University of Waterloo, Waterloo, Ontario, Canada.
(2) Piezoelectric sensor for foot pressure measurement Sudhakar Bhatt, John G. Webster, Willis J. Tompkins, and Jacqueline J, IEEE Engineering in Medicine 21 Biology Society 11th Annual International conference, (1989).
(3) Paraplegia from spinal cord injury Tzonichaki I, Kleftaras G, OTJR; Occupation, Participation, and Health-22.
(4) Detection of posture and motion by accelerometer; a validation study in ambulatory monitoring. F. Forester, M. Smeja, J. Fahrenberg, Computers in Human behavior 14, (1999).
(5) Biomechanical Design of a Powered Ankle-Foot Prosthesis Samuel K. u, Jeff Weber, and Hugh Herr.
(6) Ankle torque steadiness Brandholm T, Rose MH, Slok R, Sonne-Holm S, Jensen BR, Muscle Nerve, (2009).
(7) In shoe pressure measurement and analysis in relation to Foot Strike.
(8) Saba Naghipour, Supervisor: Dr. Robert Morley Mansfield, P.J., Newmann, D.A., (2009), Essentials of Kinesiology for the Physical Therapist Assistant, Mosby: St. Louis, MO.