Design and Development of The Control System for Powered Ankle Prosthesis

Gong Yan

A Thesis Submitted in Partial Fulfillment of the Requirements for the Degree of Master of Engineering in Electrical Engineering

Prince of Songkla University

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Major Program Electrical Engineering

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ABSTRACT

The thesis presents the power ankle walking control system based on Flexi-Force sensors for amputee patient. The system can be used for the level-ground and upslope walking for artificial power ankle. Phase measurement is a main limitation in power ankle control system because the traditional control methods was complex mechanical systems and highly complicate computation systems. This paper aims to develop a straightforward phase control system using two Flexi-Force sensors, SR1 and SR2, SR1 installed under the fore foot and sensor SR2 installed under the heel. Two signals are used for walking phase classification. Sensor SR1 and SR2 will compare with threshold and classify the walking phase. We design an artificial power ankle model for testing the system. The experiment with an artificial power ankle model are provide walking speed with 2 km/h. The signal from accelerometer ware used to walking pattern verification，accelerometer A1 and A2 are respectively installed to healthy foot and artificial power ankle model, using digital to analog conversion NI usb-6521 DAQ and LabVIEW 8.2 to record the walking data, compared the ankle model data with healthy foot walking data. The result shows that this method can measure the ankle phase during the walking in real-time. This method can be used to control artificial power ankle walking in level and upslope. It also can be used for design more movement type such as downhill slope or up and down stairs.

Keywords: gait analysis, control system, power ankle, Flexi-Force.

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Chapter 1
Introduction

1.1. Problems statement

From the first official global report on disability, the proportion of disabled people is rising to 15% of the world's population [1]. How to let the disabled recover the ability to walk is more important in future.

Nowadays the commercial available prosthesis is not power passive prostheses, it only can get the ability to stance and simple walking in fixed terrain, but the human ankle provides extra power during the walking. The first work in powered prostheses was proposed in1972s [2], there had many less research about powered prostheses. Samuel K [3-4] use the impedance control of the prosthesis control system, he uses a high power output DC motor, a transmission, a series spring Composition Series-Elastic Actuator (SEA). This SEA can get the torque feedback to the system for control, this prosthesis can mimic the normal human ankle walking. But this prosthesis has to control by PC [5], it cannot work by itself. Frank Sup [6-7] designed a new prosthesis with embedded system [8], it can walk independently in outdoor for 9km. Huseyin Atakan Varol [9] proposed Real-Time Intent Recognition for that prostheses [6, 7 and 10]. This prostheses control system signal is from inertial sensor. Mr.Ugrit Chammar designs a simple mechanical system for the artificial foot [11].

It can continuously work about 6000 steps, normal human is 5500 step/day. Therefore it can work all day with full battery. The disadvantages of that artificial prostheses are impedance control based on high performance computing system and high complicated mechanical structure. Inertial sensor is difficult to make sure the walking phase in real time, it just estimate movement trend at the beginning of the walking phase, cannot measure every walking phase during the walking.

In this thesis ware developed the control system method based Flexi-Force sensor is developed. It can accurately estimate the ankle phase during the walking in real-time, and the signal can used to control artificial power ankle walking in level and upslope.

1.2. Literature review

In 2009, Samuel K. Au [3-4] designed a spring Powered Ankle–Foot Prosthesis. He combine a high power output DC motor, ball screw transmission, spring, and leaf-spring prosthetic foot from a series-elastic actuator (SEA). This SEA provides extra power during the swing phase (Fig.1.1).

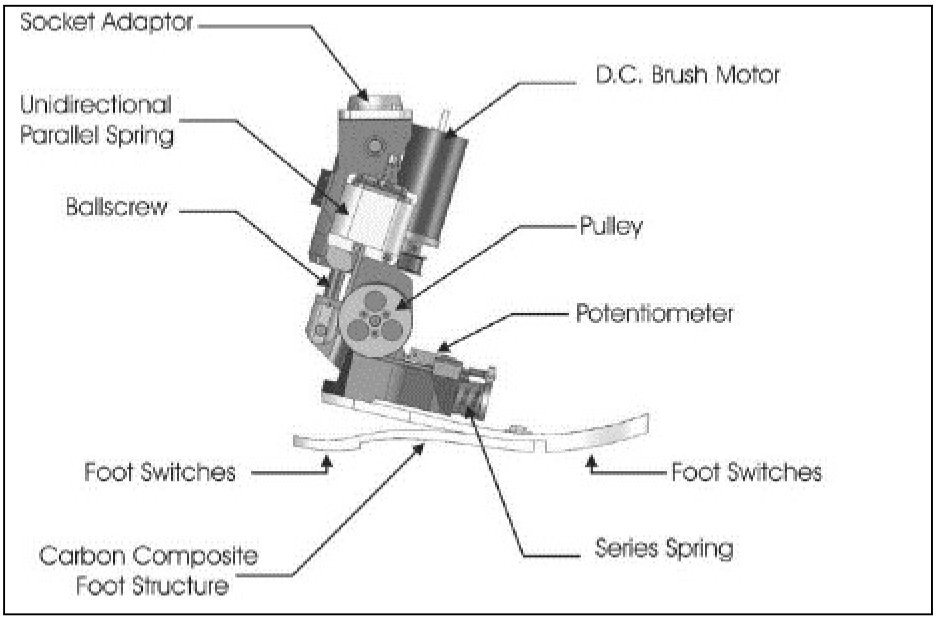


Fig.1.1 Series-elastic actuator (SEA) [4]

The control system (Fig.1.2) consists of two parts; low level servo controllers and a finite-state machine. The low level servo controllers have three types: 1) a high performance torque controller, 2) an impedance controller and 3) a position controller. The torque controller was designed to provide the offset torque, an impedance controller was designed to modulate the output impedance of the SEA. The position controller was proposed to control the balanced position of the foot during swing.

The finite-state machine has two parts: a state identification and a state control. State identification was used to identify the current state of the prosthesis and state control was used to execute the predefined control procedure for a given state.

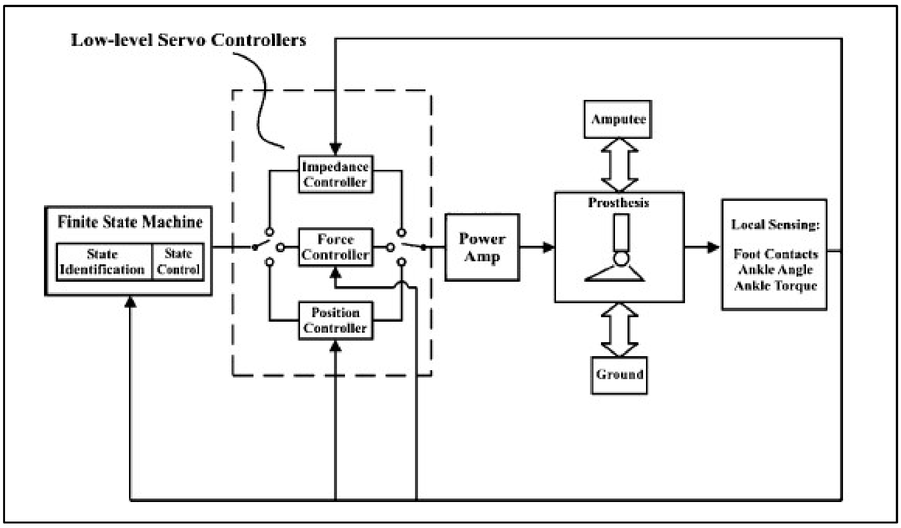


Fig.1.2 Overall control architecture of the prosthesis [4]

Fig.1.3 shows the schematics of the computer system [5]. It contains an onboard computer PC104 with a data acquisition card, power supply and motor amplifier. The system was powered by a 48 V, 4000 mAh with Li-Polymer battery pack.

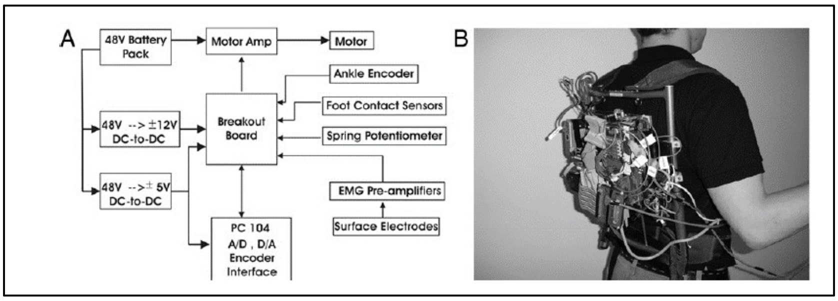


Fig.1.3 Computing System [5]

In summary, this prosthesis can give the ability to walking with 1.45m/s. But the system processing is complex that cannot use for normal walking and it also cannot walk in a slope.

In 2009 Franksup [6-7] designed a new powered knee and ankle prosthesis (Fig.1.4).

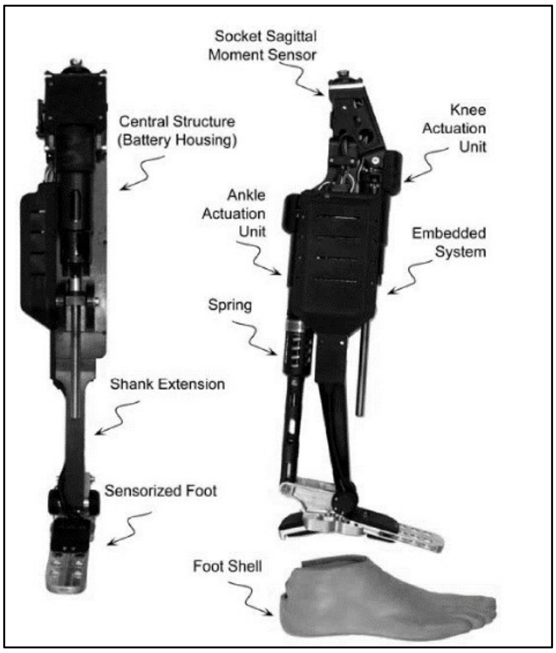


Fig.1.4 Self-contained powered knee and ankle transfemoral prosthesis [8]

This prosthesis includes the batteries, embedded system, foot sensor and moment sensor. This prosthesis can work individually in outdoor. The powered prosthesis provides a range of 12.2 km of level walking, 9.2 km of 5^0 upslope walking, and 7.7

km of 10° upslope walking between battery charges

Fig.1.5 [8] is the embedded system powered by a battery with 29.6V and 4000mAh capacity. The embedded system consists of signal processing, power supply, power electronics, communications, and computation modules.

The control system of the prosthesis consists of three levels (Fig.1.6) [8]; high-level supervisory controller, middle-level controller and low-level controllers. High-level controller is the intent recognizer, that infers the user's intent and change to the middle-level controller. The middle-level controller is developed for each activity mode. For example: standing, walking and sitting uses the finite-state machine to modulate the impedance of joints and generate torque on the different phase of the gait. The low-level controllers are the closed-loop joint torque controllers, it used to compensate the knee and ankle joint torque because the torque is consumed by friction.

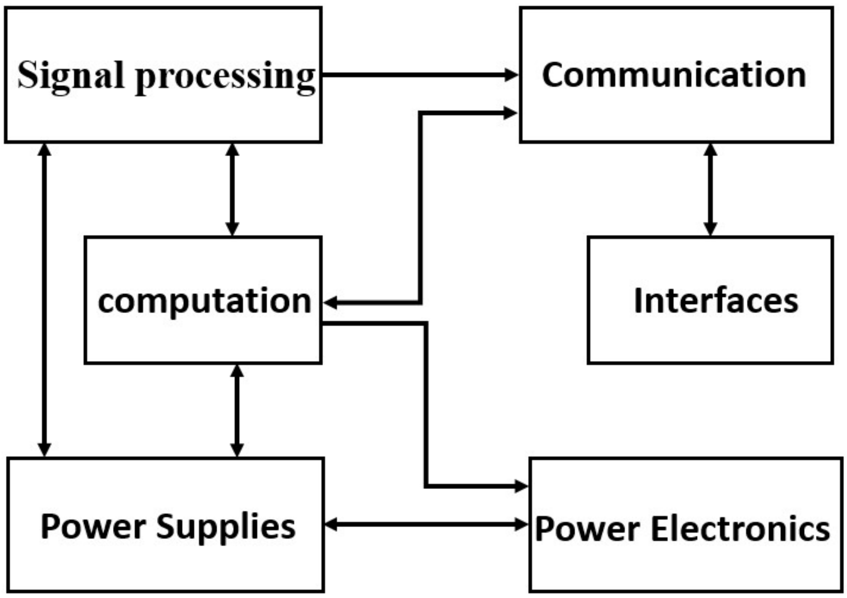


Fig.1.5 Embedded system framework [8]

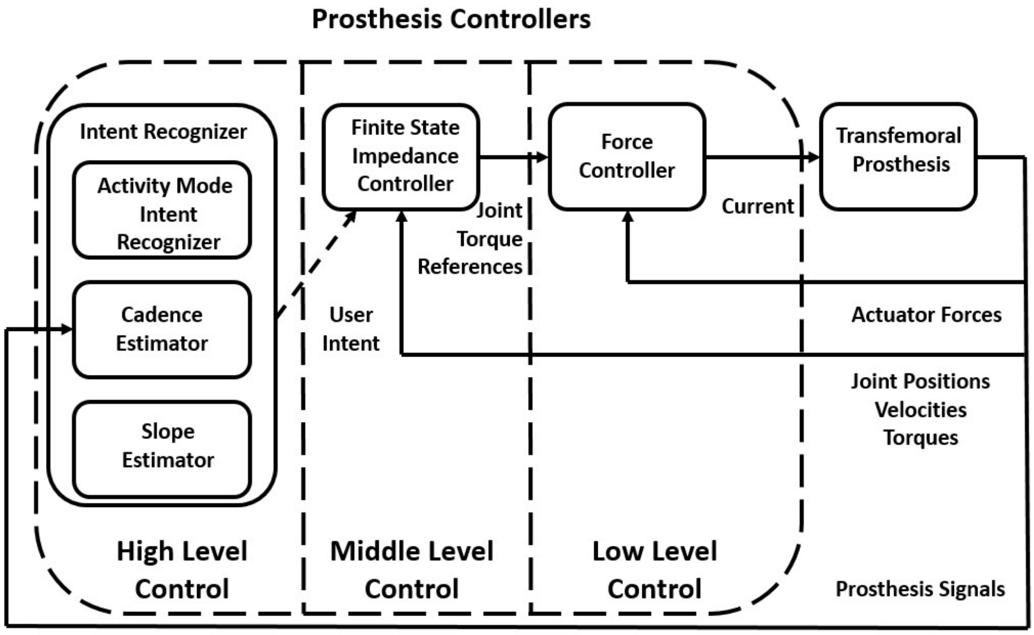


Fig.1.6 Complete control architecture [8]

This prosthesis walking in level-ground and upslope, Fig.1.7 is a threshold switching for upslope walking [10]. Based on the three-axis accelerometer (Analog Devices, ADXL330) data.

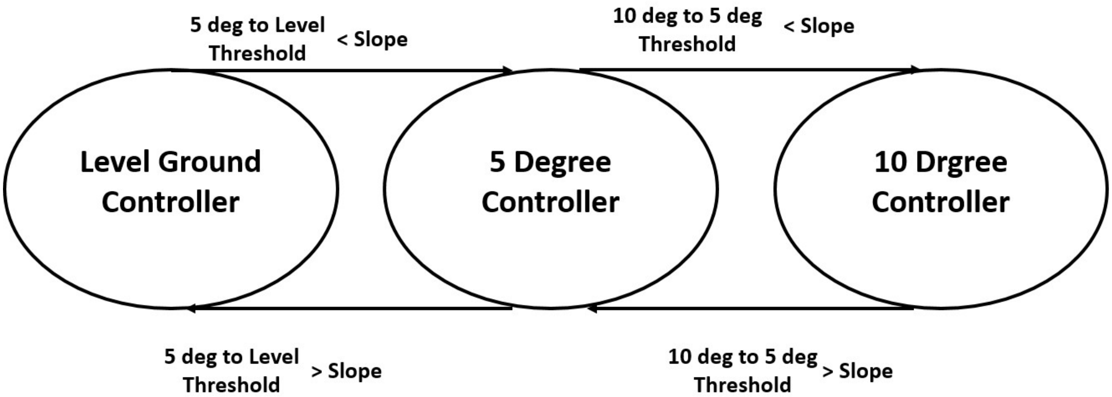


Fig.1.7 Threshold switching between joint impedance parameter sets for upslope walking [10]

In summary, this prosthesis walking speeds are about 5.3, 4.7, and 4.2 km/h (for level, 5° and 10° slopes). The powered prosthesis provides for a range of 12.2 km of level walking, 9.2km for upslope walking (5°) and 7.7 km for upslope walking (10°) with a fully charge battery. But during the walking, the walking speed was fixed.

Fig.1.8 Artificial foot [11]

Mr.Ugrit Chammar [11] designed a new artificial foot (Fig.1.8). This foot used proportional control system and used the accelerometer for the feedback system. This foot battery size is 2200 mAh and it can work continuously about 6000 step (for people 5000-5500 step/day), so it can work for one day by human. The walking speed is 1m/s (normal walking about 1.5km/h) and that artificial foot only can work in ground level.

Table1: Summary

Method	Advantage	disadvantage
Recognition of an upslope with the laser range finder[13]	Faster and high accurate slope aprediction the slopes	Calculation too Complex and too expensive
Impedance control[3-10]	Impedance control fast and accurate feedback in every walking phase	1: Complex mechanical system. 2:Complex computing systems need more power 3:Walking cadence and walking slope are fixed
Three-axis accelerometer[14]	Fast slope estimation	Complex calculation High complexity calculation procedure
Force sensor control[12]	Simple calculation	Slope response time only use in the slow motion

1.3 Objective

1 To develop the artificial foot level walking speed up to 2km/h.

2 To develop the artificial foot working period more than 12 hour with people in level-ground.

3 To develop the artificial foot in level-ground and in upslope.

1.4 Thesis scope

1. Investigation of the ankle power prosthesis in level-ground control system.

2. Design and development of the upslope control system with software simulation (Labviews).

3. The signals from accelerometers of the walking using the artificial foot are compared with the signals of the walking using the normal foot

4. Verification with the reference control system for the 5^{o}, 10^{o} and 15^{o}.

Chapter 2

Theory

2.1 Healthy walking

For normal walking (Fig.2.1), we can found that the human ankle walking cycle is defined as beginning with the heel strike of one foot and ending at the next heel strike of the same foot. In all cycle the stance phase is about 60% gait cycle and the swing phase is about 40% gait cycle.

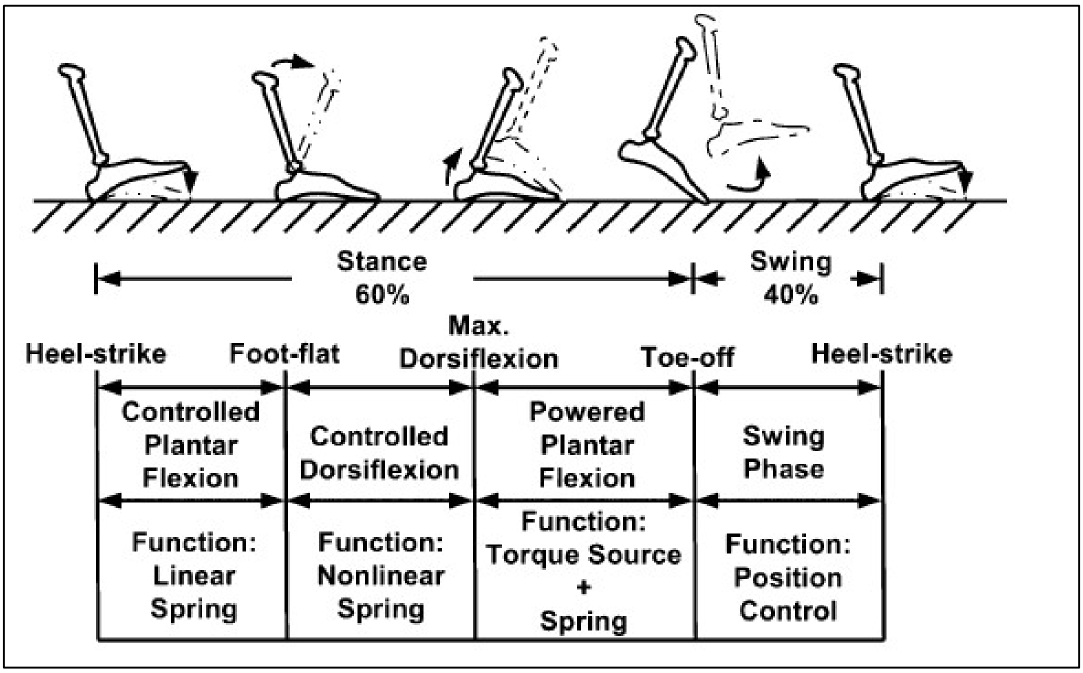


Fig.2.1 Normal human ankle biomechanics for level-ground walking [3]

The stance phase is about 60% gait cycle (Fig.2.1). It includes three phases, the first it is controlled plantar flexion , the second it is controlled dorsiflexion and the third is powered plantar flexion .The controlled plantar flexion begins at heel-strike and ends at foot-flat. The ankle angle is from 0^o to -7^o (Fig.2.2). The controlled dorsiflexion begins at foot-flat and until the ankle reaches a max dorsiflexion (from the -7^o to 10^o). The powered plantar flexion powers plantar flexion begins after controlled dorsiflexion and ends at the toe-off, and the ankle angle is from 0^o to 10^o.

The swing phase is about 40% gait cycle (Fig.2.1) starts at toe-off and ends at heel-strike. It represents when the foot is off the ground during the gait cycle. During this phase the ankle angle can be approximate stationary (Fig.2.2).

In summary, the ankle angle range is between -7° to 10° and the ankle need to provide extra power for push-off during powered plantar flexion w≈0.13j/kg. For 100kg weight people work done is about 13J/step. For normal walking 5000-5500 step/day [11], work done W=71500J/day. In this thesis we choose 12V 8000mAh battery power, this battery energy as follow:

$$Q= UIT \tag{2-1}$$

From Equation (2-1) Q is the joule, U is the battery voltage, I is the battery working current and T is the time for working. we can know battery energy is about 345600J, very suitable in this thesis.

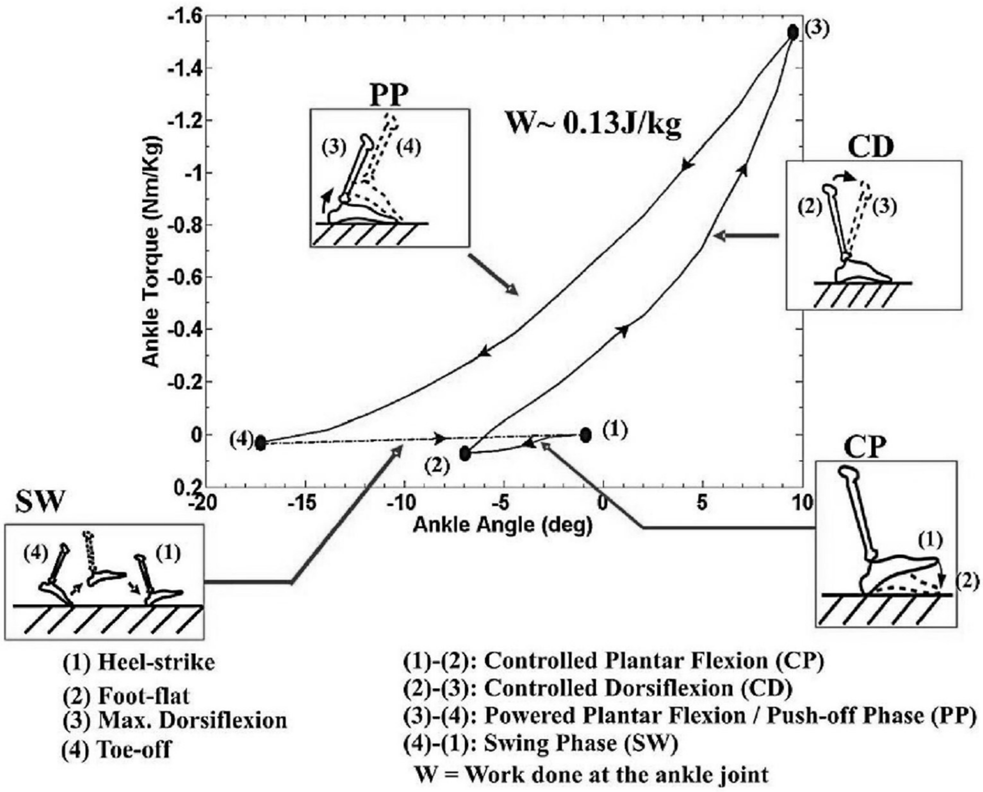


Fig.2.2 Ankle torque-angle during stance and the swing phase [3]

2.2. Control method

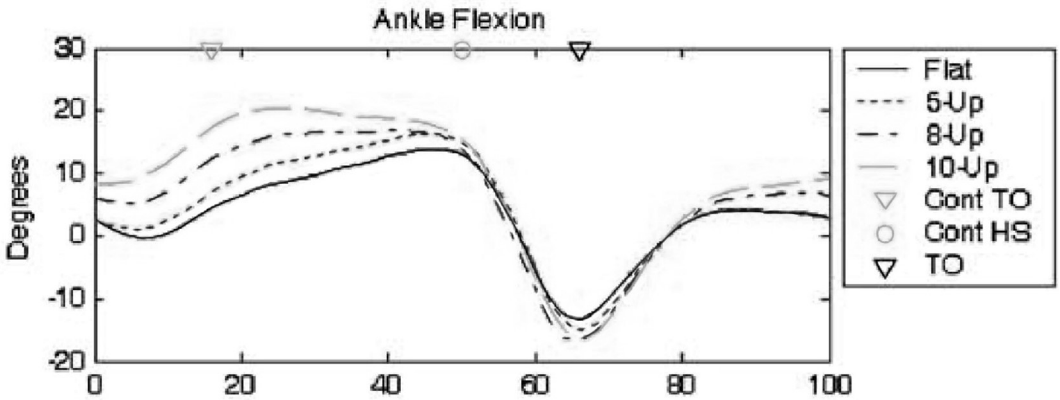


Fig.2.3 the correlation between walking level and upslope degree [15]

From A.S. McIntosh [15] (Fig.2.3), he research about the ankle joint during the level, 5°, 8°, and 10° walking. Form this result, we can find the level walking and up slopes are similar. The different is the degree of balance and intention recognition. Definition of clockwise rotation will be denoted by a positive and reverse travel as negative, 0° is the base when the leg and foot is vertical and threshold is parameter depending on body weight. In this thesis, data from gait dynamics [15] will use for parameter in control system feedback, parameter as follow:

1. Level-ground: heel-strike start at 0° ends at 10°, max-dorsiflexion was -10°.

2. 5^0 upslope: heel-strike start at 0° ends at 15, max-dorsiflexion was -10°.

3. 10° upslope: heel-strike start at 10° ends at 20°, max-dorsiflexion was -10°.

Accelerometers can estimate the slope in stance phase and gyro can estimate the slop in swing phase. The connection of two sensor, the slop can be estimate in any time, but the calculation will be too complex [14].

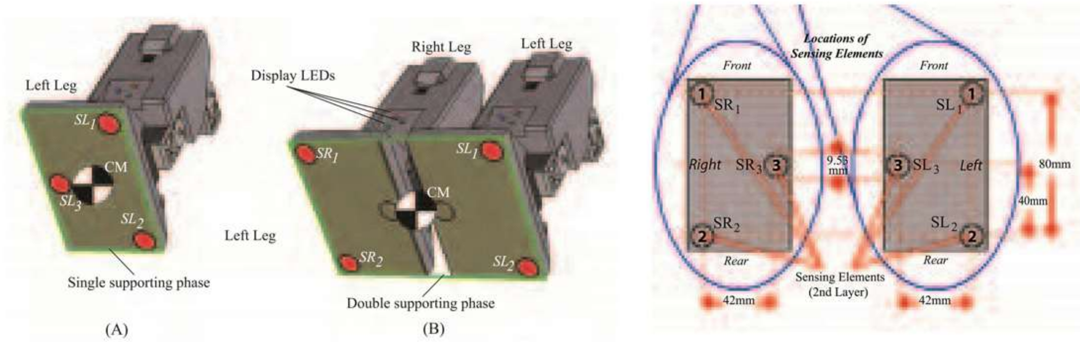
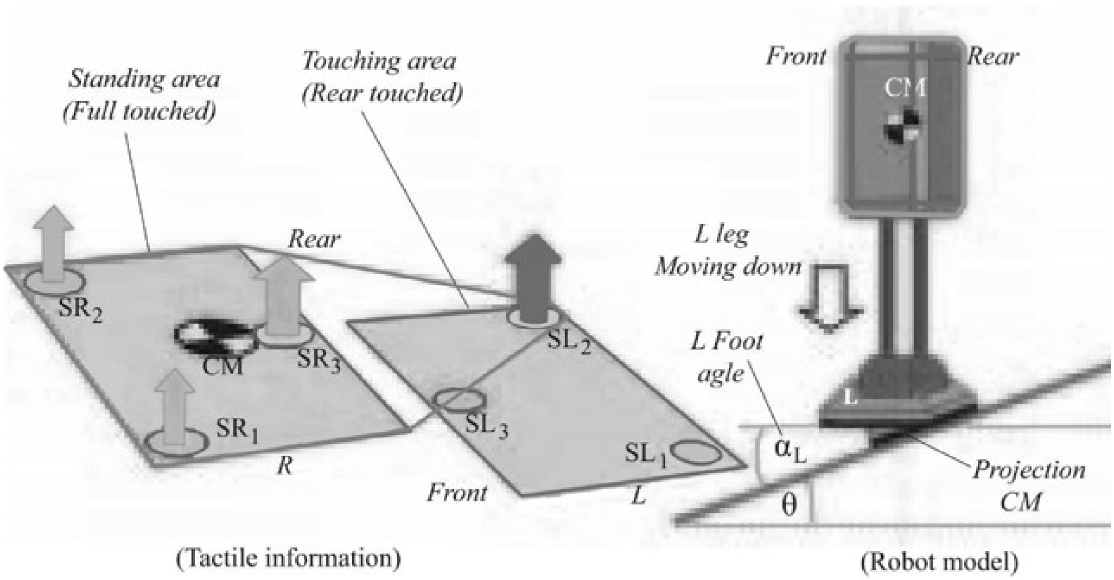
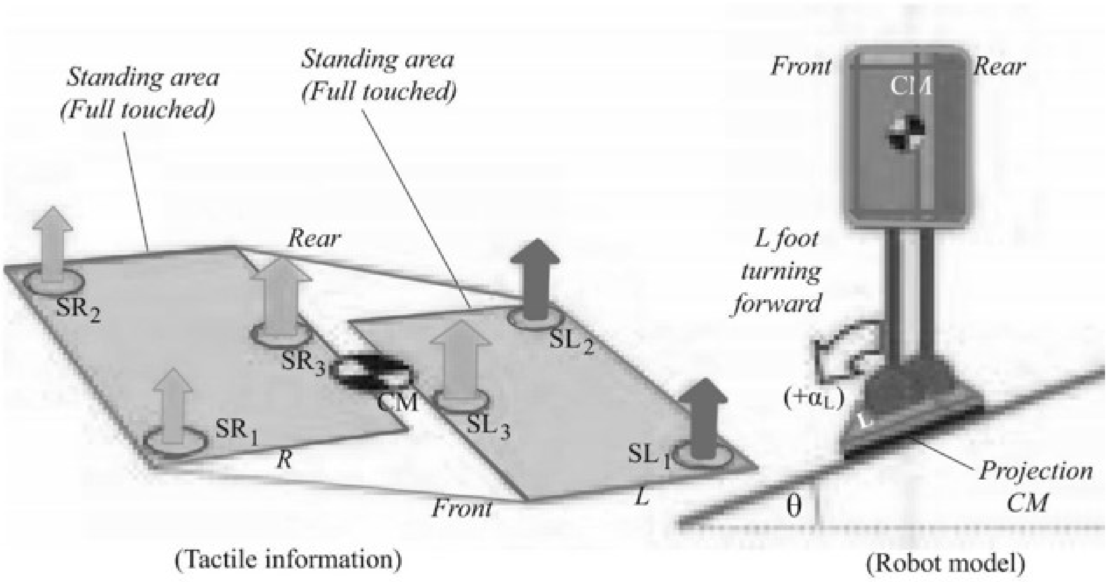


Fig.2.4 Sensors positions [12]

This method is from K. Suwanratchatamanee (Fig.2.4), force sensor is used to measure the slope, already inform the three sensing elements are fixed to make triangular position on each foot (the average allocation the force).



(a)



(b)

Fig.2.5 working condition (a):SR1>SR2, (b):SR1=SR2 [12]

First step is the intention recognition when the foot touch the ground, the sensor output changed in this case the SR1>SR2 (Fig.2.5 (a)), the system will know the ground degree already changed, and the turns the ankle degree until SR1=SR2 (Fig.2.5 (b)), because when SR1=SR2, we can found the body already balance, so the system can keep the ankle degree and recording the degree data. This method advantage is a simple calculation, unnecessary to estimation of ground slope, but defect is it use for dynamic case when the motion is slow in the real time. The SR3 was used to keep the robot balance during the robot switch the two leg.

2.3. Level and upslope walking with Flexi-force sensor

2.3.1 Flexi-force sensor

Fig.2.6 Flexi-force sensor Model A201 [16]

Flexi-force (Fig.2.6) is an ultra-thin force sensor has the following advantages: thickness only 0.208mm, low response time less than 5 microseconds, sensing area is 9.53mm diameter, suitable in this thesis.

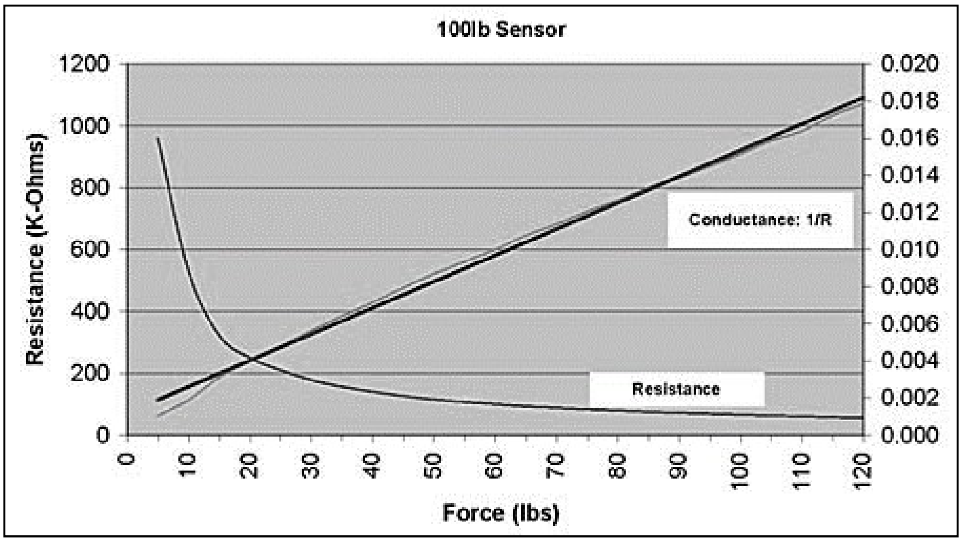


Fig.2.7 The relationship between force and resistance of the sensor [16]

When the force sensor is unloaded, the resistance is more than 5MΩ. When a force is applied to the sensor, this resistance decreases (Fig.2.7). This sensor can use non-mechanical structure to detect the walking phase during the walking in real-time.

This sensor used Force-to-voltage circuit, is a amplifier circuit. The circuit output is V_{out}=-V_t*(R_f/R_s). In this circuit Flexi-force sensor similar to a variable resistor. Applying a lower drive voltage and reduce the resistance of the feedback resistor can increases the range but the sensitivity will decreases (Fig.2.8).

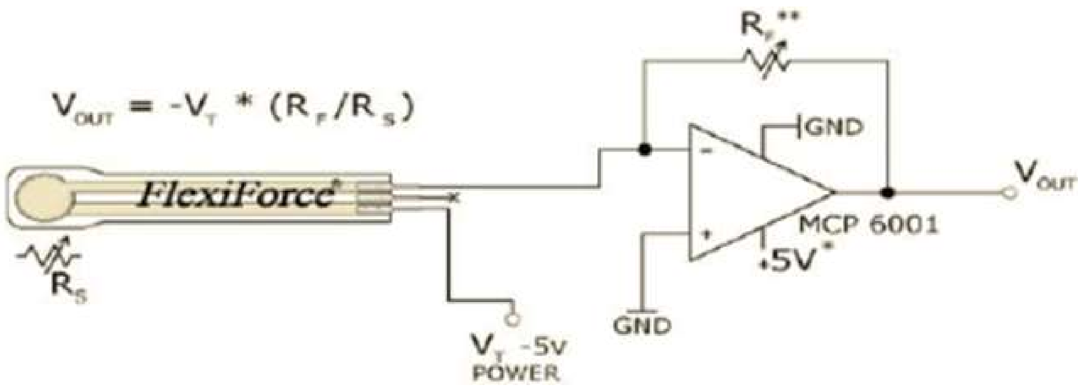


Fig.2.8 Driver Circuit [16]

The force and output voltage were linear relationship. (Fig.2.9) data was from this sensor company the force scope is 0-100(lbs). We also tests the sensor force scope from 0 to 10kg. For the variable resistance R_f, this thesis we choose 750kΩ. The result is nearly linear (Fig.2.10).

From pressure formula p=F/S, this Flexi-force sensor sensing area is 9.53mm diameter so the pressure in 10kg is about 1.367x106 PA and for normal person the sole area is about 200-250 cm^2 Calculation from pressure formula the Sole pressure no more than 3.92x104 PA, 10kg was suitable in this thesis.

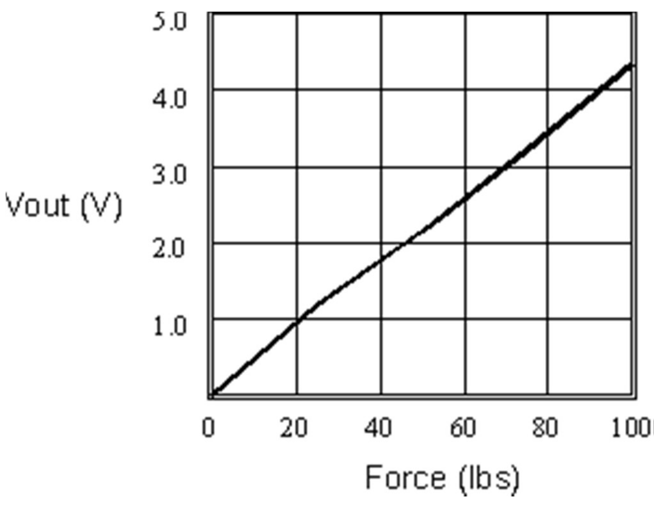


Fig.2.9 The correlation between force and output voltage [16]

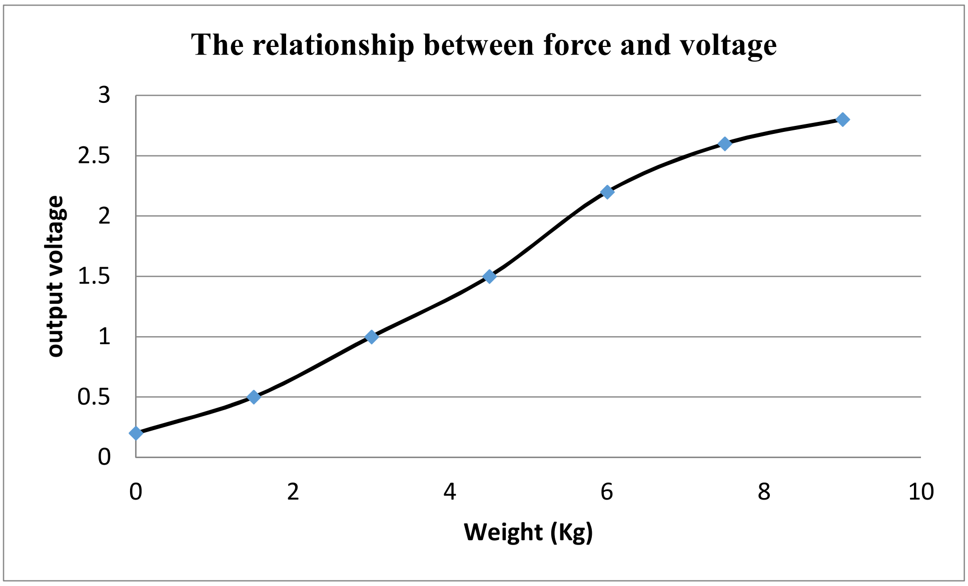


Fig.2.10 The comparison result between force and output voltage

2.3.2 Level walking controller method with Flexi-force sensor

The walking phase can be simple and accurate expression by two Flexi-force sensor SR1 and SR2. The key question for the power ankle control system is an ankle angle control. A. S. McIntosh measured the ankle angle during the walking on level, 5° and 10° [15].

In this thesis, we only use the level-walking data. Definition of clockwise rotation will be denoted by a positive and reverse travel as negative, 0° is the base when the leg and foot is vertical and threshold is parameter depending on body weight (Fig.2.11). The controller method is described as follows:

a. Standing to swing phase: system started in the standing and wait for sensor signal. When the foot is off the ground, sensor SR1 and SR2 signal equal to zero, in this time reset the ankle angle to zero.

b. Swing phase to max dorsiflexion phase : In this step when the heel touches the ground, SR2 > threshold and SR1 = 0, control the ankle angle from 0° to 10° and reverse travel to -7°, delay before the reverse travel for protecting the DC motor drive circuit

c. Powered plantar flexion: In this step the heel is off the ground and the toe still on the ground, SR1 > threshold and SR2 < threshold. Reset the ankle angle to 0° and wait for a next gait cycle.

d. In this thesis the threshold we choose 0.5V. Because the threshold was smaller and the system response time was shorter.

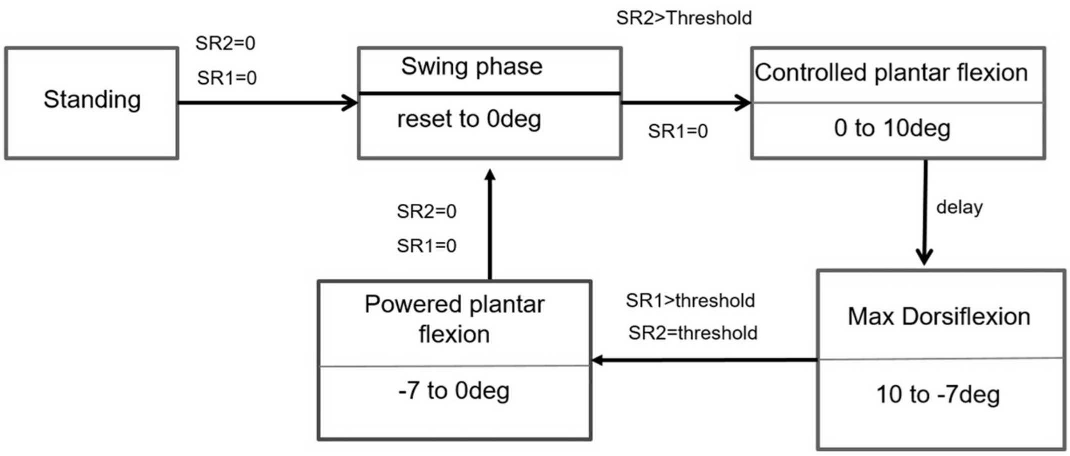


Fig.2.11 walking controller with flexi-force signal

2.3.3 Upslope walking

The control method (Fig.2.11) not only can be used to level walking but also can be used to upslope walking. From result (Fig.2.3) we can know the level walking and up slopes are similar. The differences are the degree of balance and intention recognition. For upslope walking, the control system just changing the parameter data from level walking to 5^{o} or 10^{o} upslope walking data and testing the walking result. The upslope data is from reference [15].

2.3.4 System feedback with encoder signal

From (Fig.2.11), this system feedback is degree of the motor. In this thesis we choose encoder sensor to record the degree position. Micro controller DSPIC has Quadrature encoders interface mode, that mode used in position and speed detection of rotating motion systems (Fig.2.12). QEA and QEB are encoder output signal, quadrature decoder captures the phase signals and index pulse and converts the information into a numeric count of the position pulses (POSCNT register). Generally, the count will increment when the shaft is rotating one direction and decrement when the shaft is rotating in the other direction.

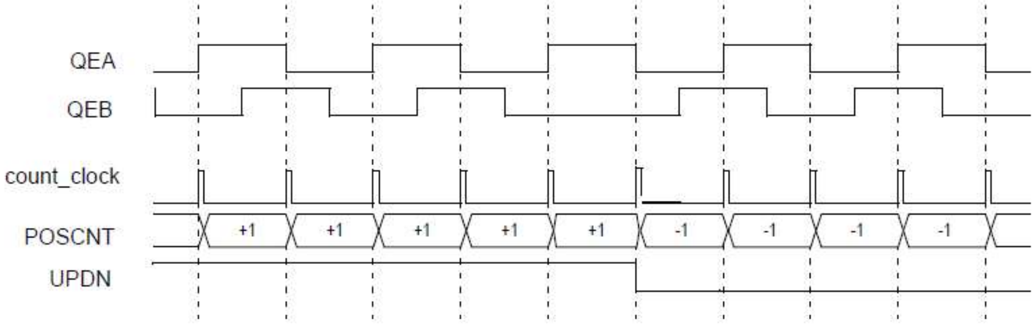


Fig.2.12 Quadrature Decoder Signals in 2X Mode

Chapter 3
Hardware design

3.1. System design

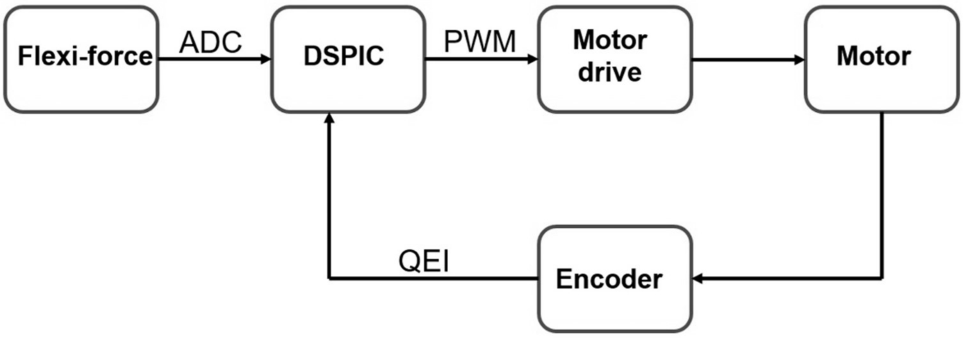


Fig.3.1 Control system for power ankle model

For the control system design, this system consists of five parts: signal collection, Micro controller, motor drive motor and encoder feedback (Fig.3.1). The function with each part as follows:

1. Micro controller: This system micro controller support three function Analog-to-Digital, motor controller and encoder signal processing, in this thesis use DSPIC as micro controller.

2. Signal collection: The Flexi-force sensor signal from drive circuit was analog signal, use Analog-to-Digital Converter (ADC) transform to digital signal and connect with DSPIC.

3. Motor drive: the motor drive circuit from (Fig.3.5) control by PWM signal, PWM motor controller function in DSPIC can control it.

4. Encoder feedback: This system is closed-loop system. The system feedback is encoder signal, DSPIC function Quadrature Encoder Interface Module (QEI) can easy to processing encoder signal.

3.2 Current design

In the current source design, the basic is H-bridge electronic circuit, that enables a voltage to be applied across a load in either direction. These circuits are often used in robotics and other applications to allow DC motors to run forwards and backwards (Fig.3.2). In this current switch S1 and S2 or S3 and S4 cannot be open in the same time, otherwise will short circuit. Switch S1 and S4 open the motor run forwards, switch S3 and S4 open the motor backwards.

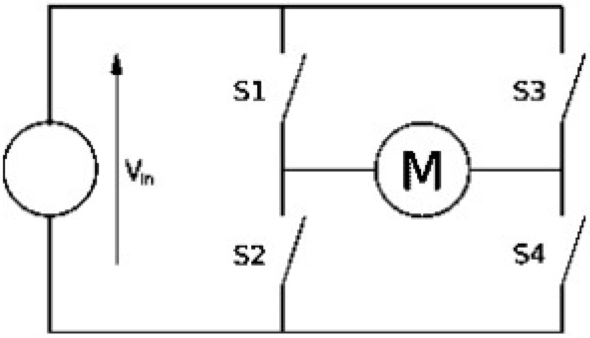


Fig.3.2 H bridge electronic circuit

The H-bridge electronic circuit have a problem is the S1 to S4 had control by human is non-automatic. Drive IC IR2110 (Appendix C) can drive the MOFSET using PWM signal, it can use micro controller to control it. In this thesis we used two IR2110 drive IC and four IRFD220 MOSFET to make the H bridge circuit. The IR2110 is a high speed power MOSFET drivers, and it also has isolated (small size) and electromagnetic isolation (high speed) advantages. IRFD220 (Fig.3.3) is N-Channel enhancement mode silicon gate power field. It can control the ID (drain current) from the VGS (gate source voltage). IRFD220 will replace the switch S1 to S4 in H bridge circuit and the G point will connect with the IR2110.

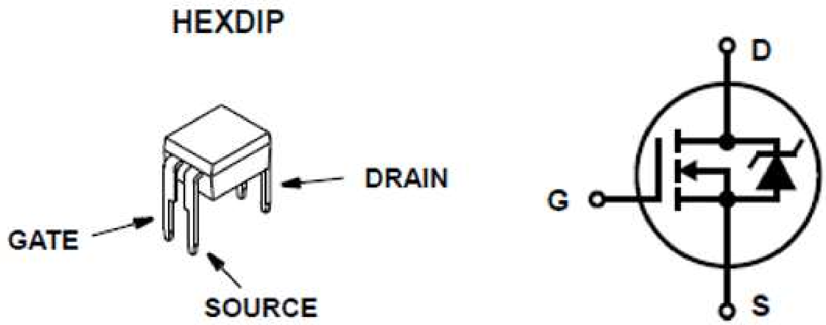


Fig.3.3 IRFD220 MOSFET

The modified H bridge circuit (Fig.3.4) include two drive IC and four MOSFET, in this circuit it control by two Channel complementary PWM signal (p1),in this case the PWM signal choose 30kHZ testing in 12V gear motor(Fig.3.7).12V is power for

IR2110 and V motor is power for motor.

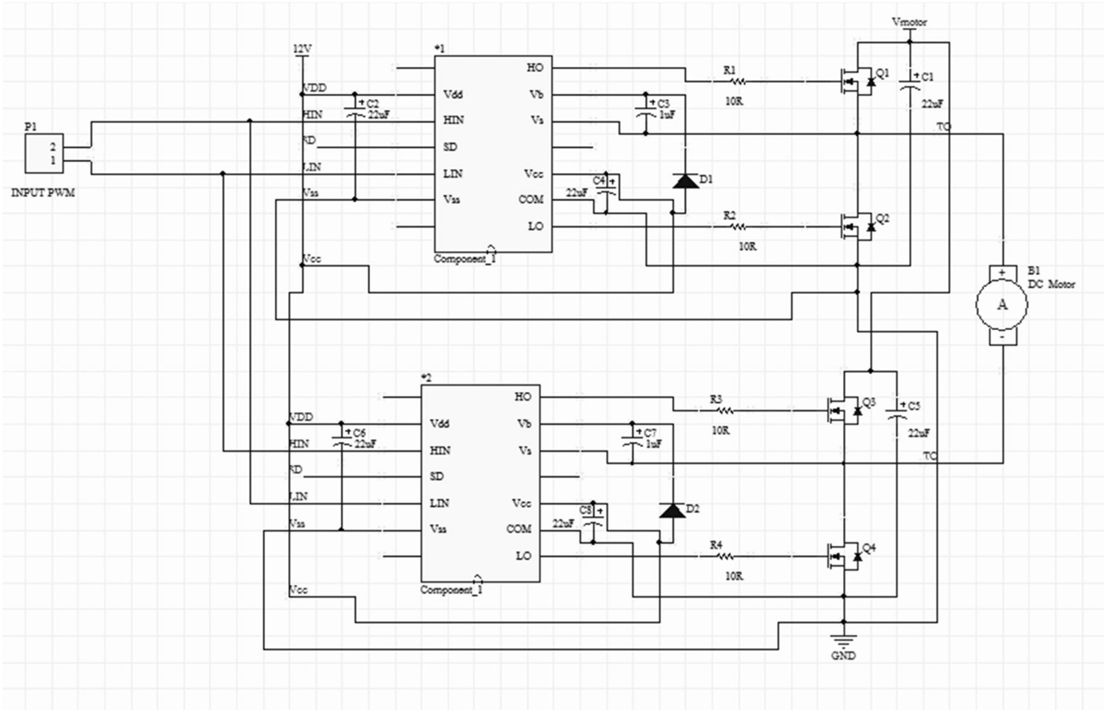


Fig.3.4 H bridge circuit design

The motor speed and direction control by PWM duty cycle, 50% duty cycle is critical point. Duty cycle less than 50% motor direction is positive and duty cycle more than 50% motor direction is negative. Motor driver test result in (Fig.3.5), X-axis was duty cycle Y-axis was motor speed frequency, signal from Encoder (100PPR).

From this result we can know the motor speed and duty cycle change was linear. This H bridge drive circuit was working.

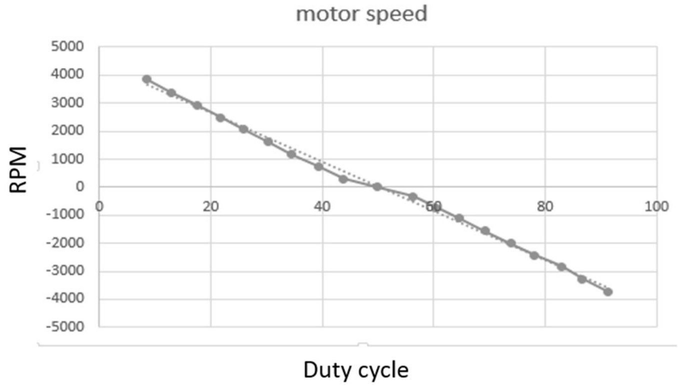


Fig.3.5 motor driver testing result

About the motor in this thesis we choose 12V DC gear motor (Fig.3.6).this motor full power working speed is 5500RPM, the reduction ratio is 1:100.

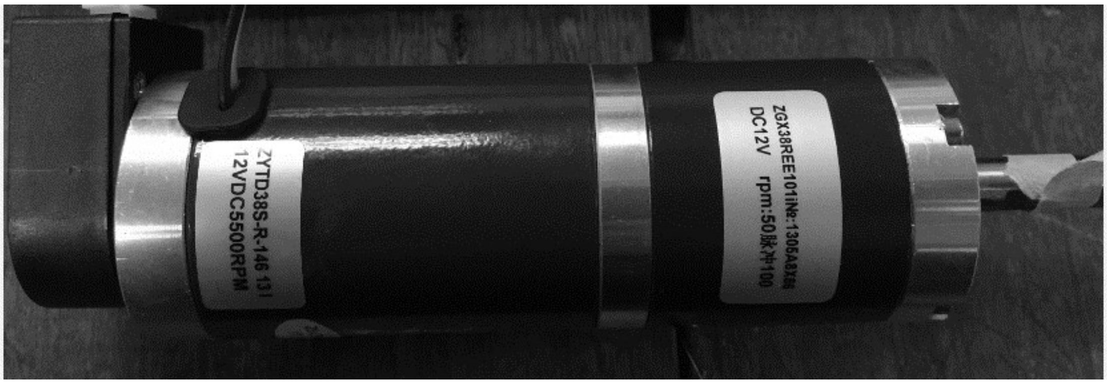


Fig.3.6 12V gear motor

For the system feedback in this thesis we choose Incremental Rotary Encoder. The body sizes between 30mm and 40mm diameter, power by 12-30V DC, Output signals include 10-30V DC, encoder is about 100RPM/rad. This encoder have three channel signal output: A, B and Z. channel A and B offset from each other by 90 electrical degrees. In one direction the leading edge of channel A will be before the leading edge of channel B. And in the opposite direction channel B will lead channel A (Fig.3.7).

But in this thesis the output signal voltage was too high for Micro controller, so in this thesis we used voltage divider circuit reduced the outpour signal voltage from 10V to 5V.

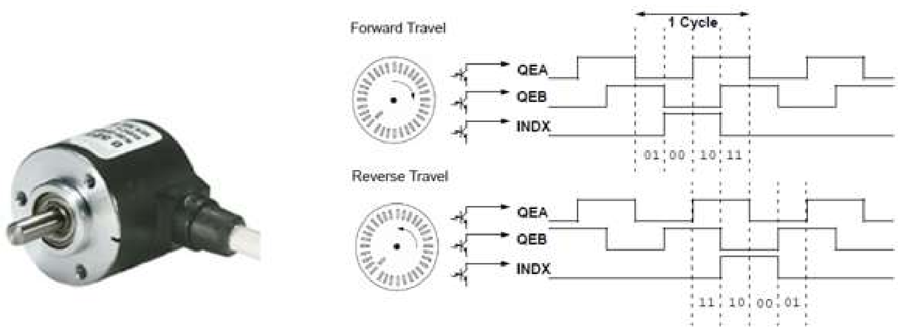
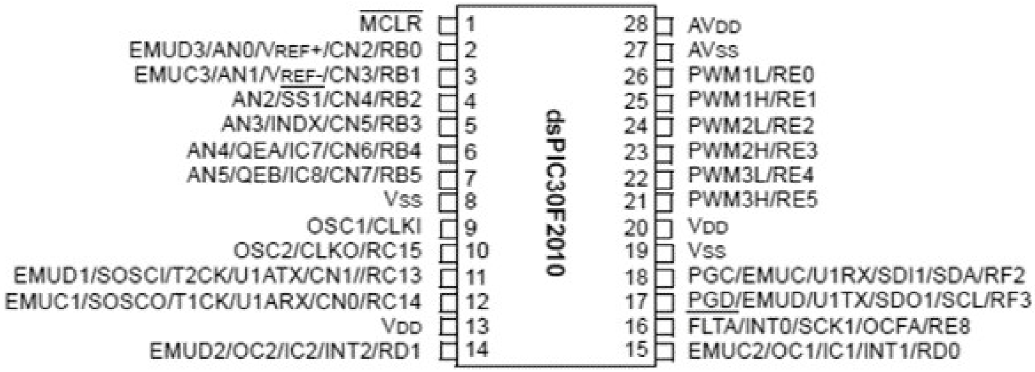


Fig.3.7 12V Incremental Rotary Encoder

Dspic30f2010 was the Micro controller for this system, is a High-Performance, 16-bit Digital Signal Controllers, performance includes: 512 bytes on-chip data RAM, DC to 40 MHz external clock input, 10-bit Analog-to-Digital Converter (ADC), Quadrature Encoder Interface Module(QEI) and Motor Control PWM Module ,this controller power by 3-5V(Fig.3.8). For the Micro controller pin connection, pin 25 and 26 connect with driver circuit signal P1, pin 6 and 7 connect with encoder A and B phase, pin 4 and 5 connect Flexi-force sensor drive circuit signal output point.



Device	Pins	Program Mem. Bytes/ Instructions	SRAM Bytes	EEPROM Bytes	Timer 16-bit	Input Cap	Output Comp/Std PWM	Motor Control PWM	A/D 10-bit 1 Msps	QEI	UART	SPI	I²C™
dsPIC30F2010	28	12K/4K	512	1024	3	4	2	6 ch	6 ch	Yes	1	1	1

Fig.3.8 Micro controller

3.3 Gait measurement using Flexi-force

In this thesis, we uses two Flexi-force A201Sensor: SR1 and SR2 are installed under the sole (Fig.3.9). Sensor SR1 is installed under the fore foot and sensor SR2 is installed under the heel. Signals from the SR1 start with a heel-strike when the heel touches the ground and ending at the dorsiflexion when the heel rises from the ground. Sensor SR2 start with a foot-flat and ending at the toe off (Fig.3.10).

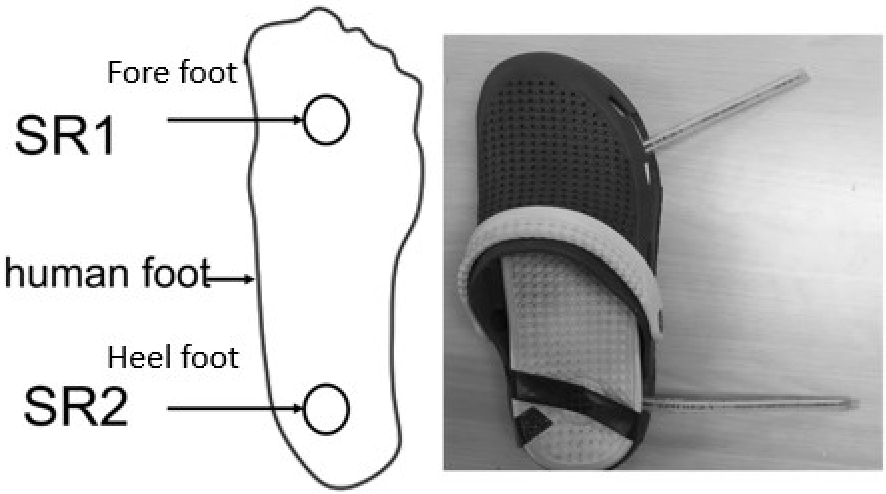


Fig.3.9 Flexi-force A201 Sensor position

The sensor shoes was tested on a 27-year-old male (1.86 m, 90 kg) healthy man .The walking data collection from DAQ IN-6009 using LabVIEW Software, sampling rate is 2000/s. Pressure on Sensor is proportional to the voltage. X axis is voltage, Y axis is time. Each step is described as follows (Fig.3.10).

a. Controlled plantar flexion begin at heel-strike and ends at the foot-flat. In this phase when heel touch ground sensor SR2 generates the signal.

b. Controlled dorsiflexion begin at the foot-flat and continues until the ankle maximum angle. In this phase, sensor SR1 generates the signal at foot-flat and SR2 stop signal at an ankle maximum angle.

c. Powered plantar flexion begin at max dorsiflexion and ends at the toe-off phase. In this phase, when the toe off the ground, the signal SR1 is reduce to zero.

d. Swing phase begin at after powered plantar flexion until the next heel-strike. In this phase the foot was off the ground, both of the sensor signal SR1 and SR2 was equal to zero.

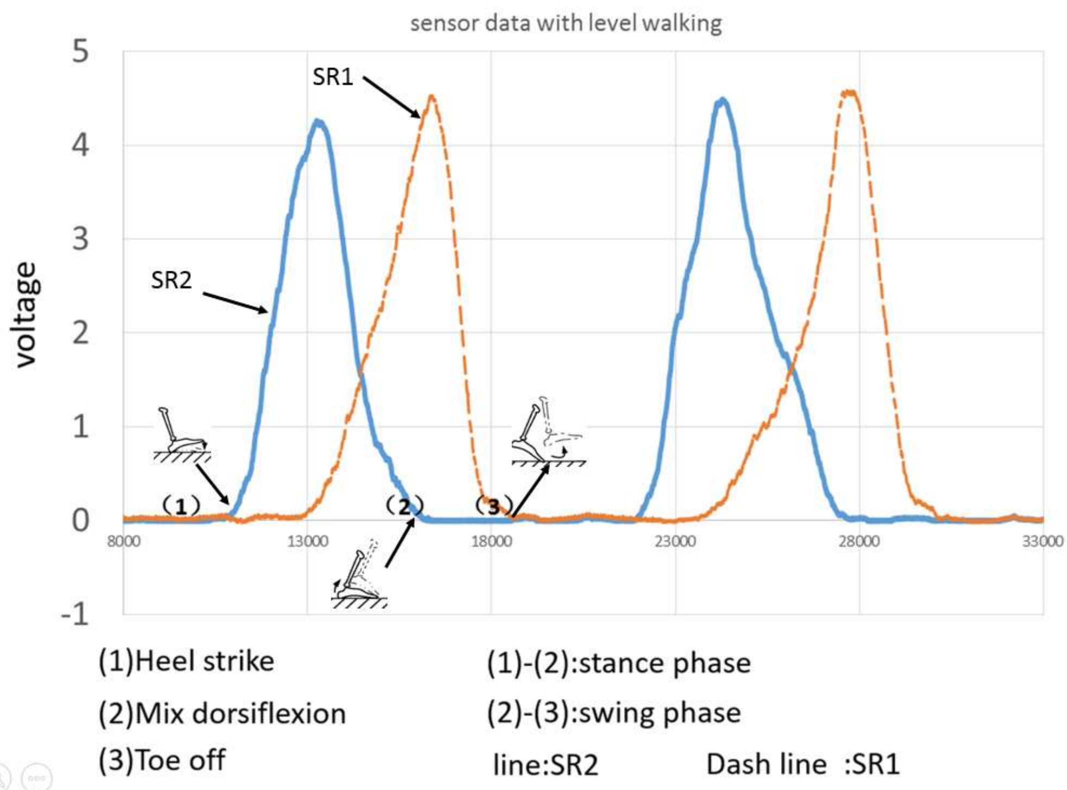


Fig.3.10 Flexi-force Sensor data with level walking

3.4 Software design

The system programming in MikroC PRO for dsPIC software use C (programming language), the flow chart show in(Fig.3.11).

For the main program was state detection(Fig.3.11(a)), analog signal from Flexiforce sensor SR1 and SR2(Fig.2.9) using micro controller ADC translate to digital signal and recorded on micro controller register, the main program will distinguish three states:

1. SR1< threshold and SR2< threshold: sensor SR1 and SR2 don't have signal, this is in the swing phase.

2. SR2> threshold and SR1< threshold: sensor SR2 just touch the ground, in the heel-strike.

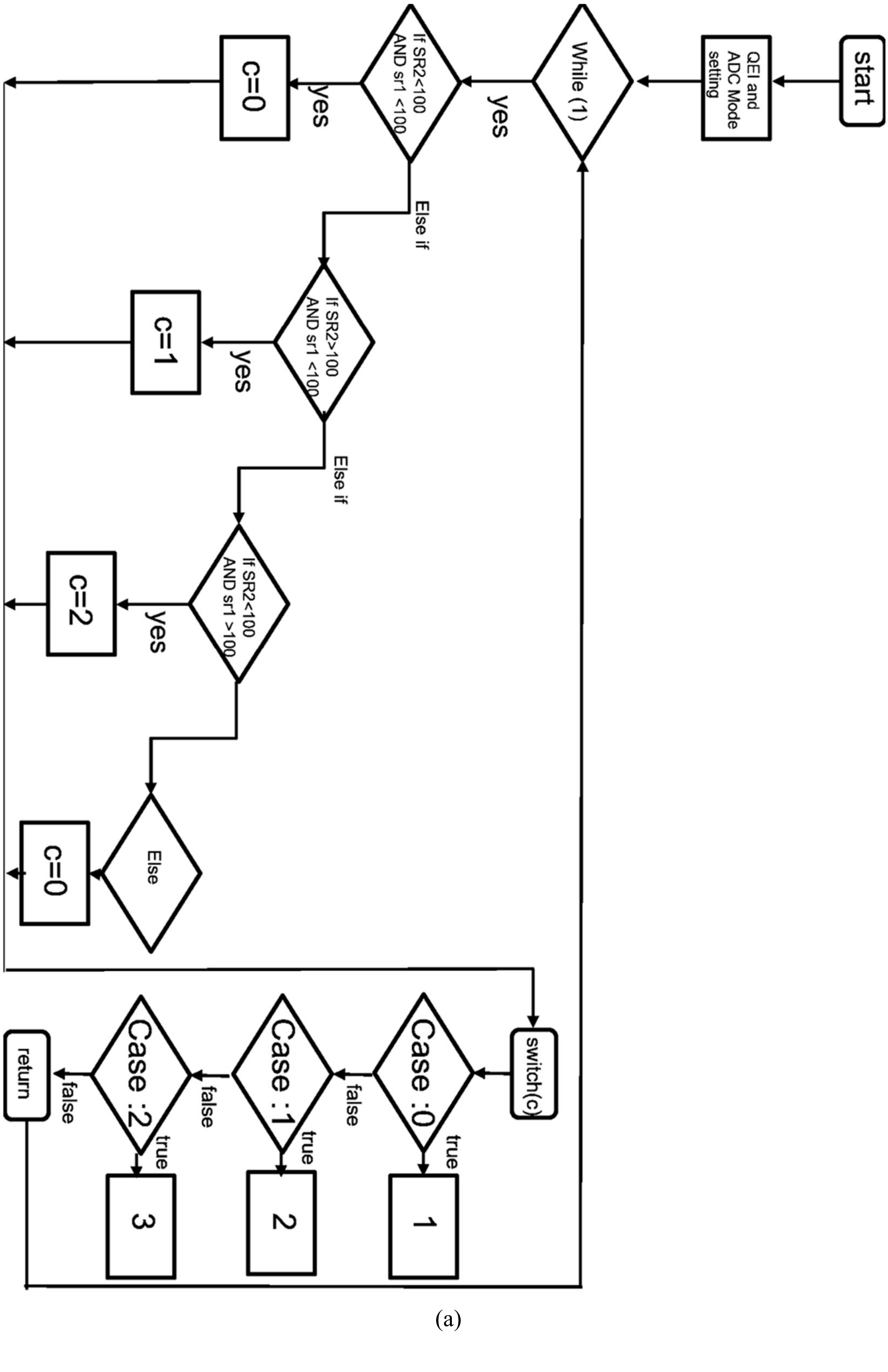
3. SR2< threshold and SR1> threshold: sensor SR1>SR2, between Max Dorsiflexion and Toe-off phase.

After State detection the main program will switch to different subroutines (Fig. 3.11(b)). Each subroutine function as follows:

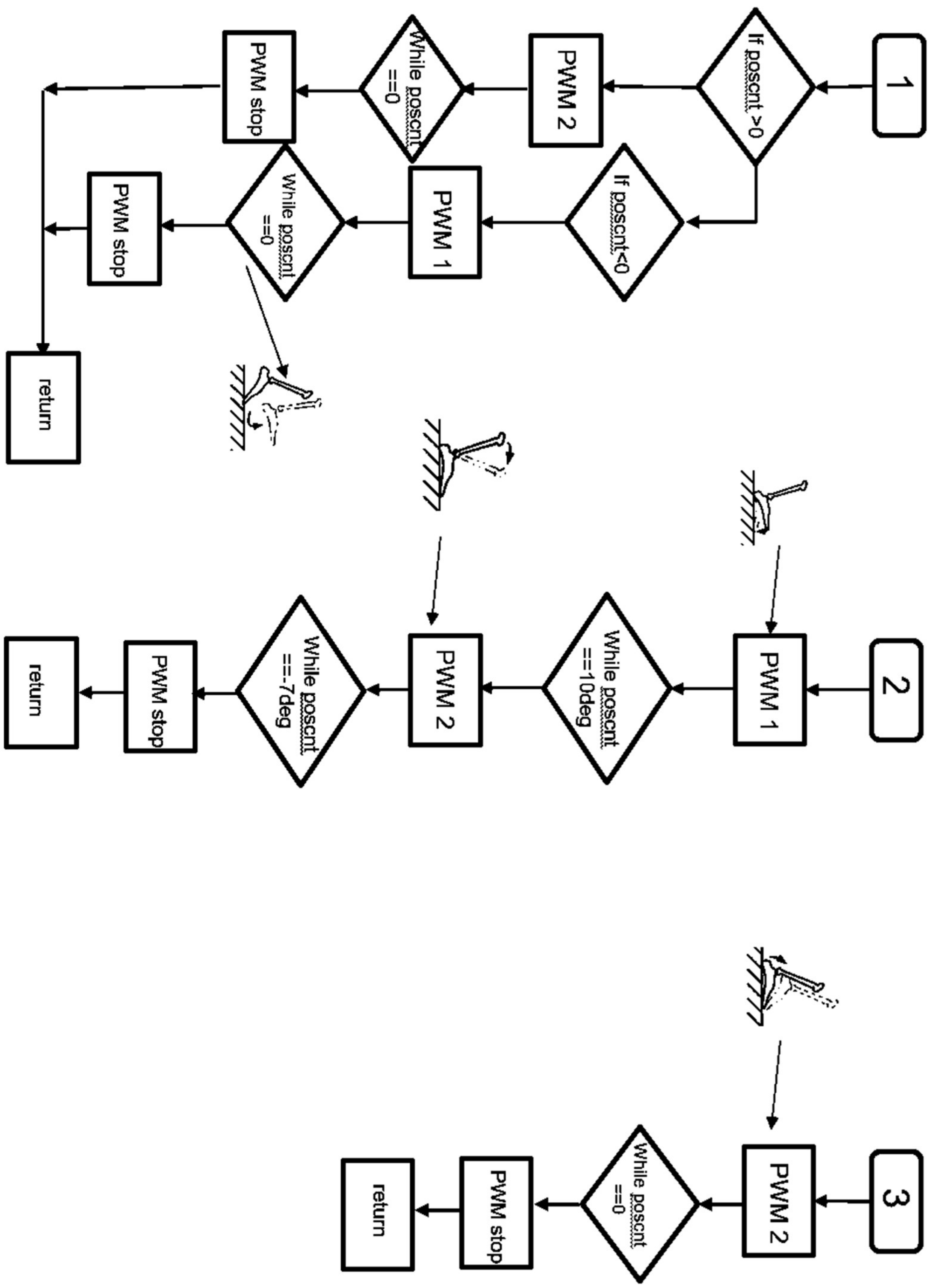
1. Program (1): In this function the program always working in swing phase, and keep the ankle degree to 0^{o}.

2. Program (2): This function is start in heel-strike and end in max-dorsiflexion, the ankle degree from 0^{o} to 10^{o} and return to -7^{o}.

3. Program (3): In this function start in max-dorsiflexion and end in Toe-off phase, the ankle degree from -7^{o} to 0^{o}.


start
QEI and ADC Mode setting
While (1)
yes
If SR2<100 AND sr1 <100
yes
c=0
Else if
If SR2>100 AND sr1 <100
yes
c=1
Else if
If SR2<100 AND sr1 >100
yes
c=2
Else
c=0
switch(c)
Case :0
true
1
false
Case :1
true
2
false
Case :2
true
3
false
return

(a)



(b)

Fig.3.11 Program flow chart (a): main program (b): subroutine

Chapter 4
Experiment and result

4.1. Artificial power ankle model design

In this article, we use plastic to make a simple power ankle model (Fig.4.1). The left one is a sensor in the shoe to collect the signal, right one is the power ankle model. The model consists of processing unit, power supply, encoder, DC-motor, and DC-motor drives. This system is powered by 12V and 700mA. The main computational element of the system is 16-bit digital signal controller dspic30f2010. Three-phase encoder can feedback the ankle angle signal to measure the ankle position. Dspic30f2010 consists of 10-bit high-speed analog to digital converter (ADC) modular and quadrature encoder interface (QEI) modular. The analog signal of the sensor can be converted to a digital signal by ADC modular, QEI modular can analyze encoder signal (Fig.3.1).

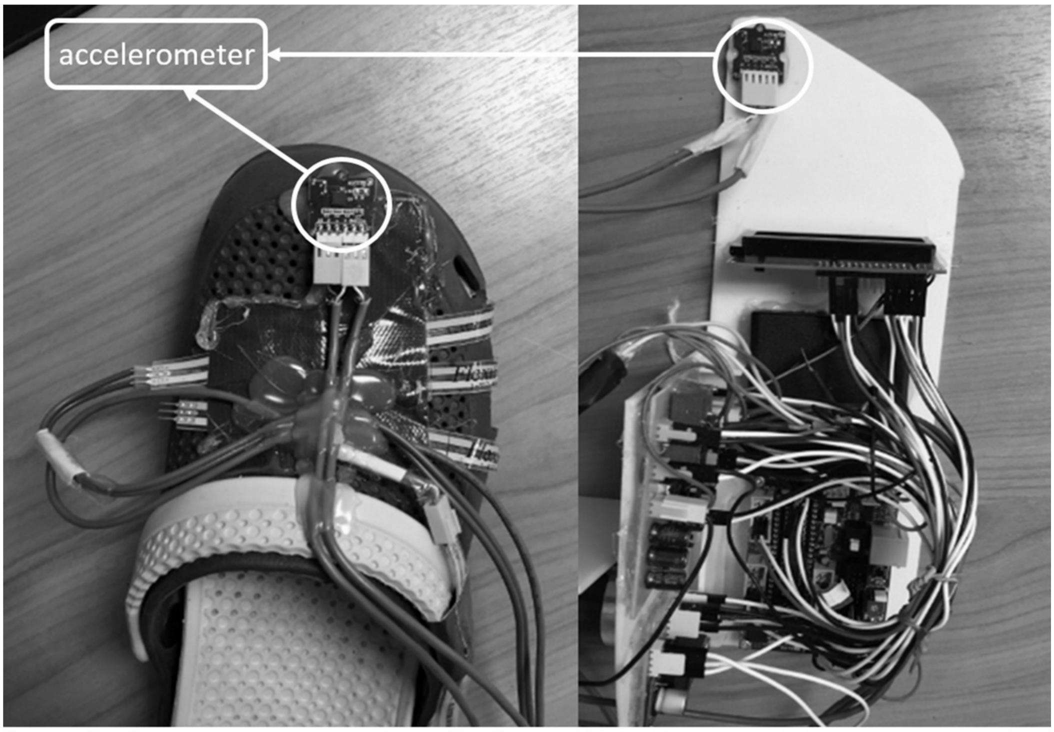


Fig.4.1 Sensor shoe and power ankle model

4.2. Experimental methods

Three axis accelerometer ADXL335 can measure three directions acceleration XYZ. Z-axis vertical to level-ground. Ankle phase can be reflected in Z-axis. Accelerometer A1 and A2 respectively are installed on Sensor shoe and power ankle model to verify work status of the control system (Fig.4.1).

The artificial power ankle is fixed on the thigh, artificial model near with the foot and vertical the leg (Fig.4.2). Sensor shoes will be worn on the same side of the foot. During the walking, accelerometer A1 and A2 will have two different signals, the two output signals were compared the similarity to control system validation.

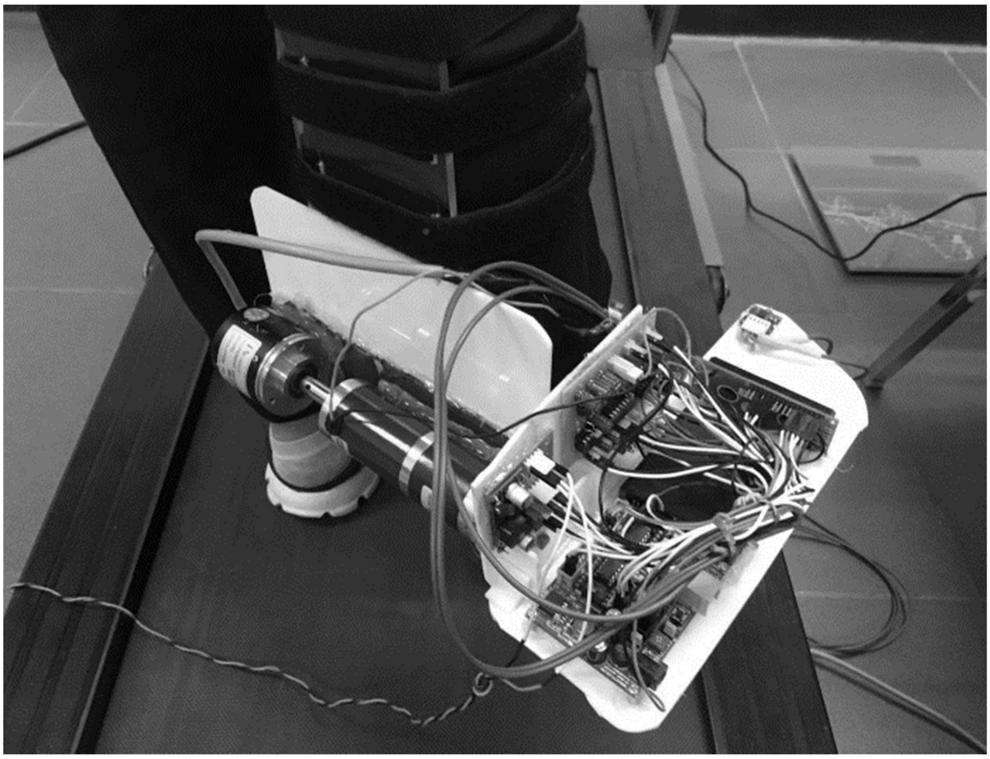


Fig.4.2 artificial model installation location

Accelerometer A1 and A2 by means of DAQ NI-USB-6251 collection, USB-6251 have 16 analog inputs (16-bit), 1.25 MS/s single-channel (1 MS/s aggregate), 2 analog outputs (16-bit, 2.8 MS/s); 24 digital I/O (8 clocked); 32-bit counters. With plug-and-play USB connectivity, these devices are simple enough for quick measurements but versatile enough for more complex measurement applications (Fig.4.3). DAQ by means of USB to PC computer and use software Labview to analysis and processing. In order to better prove that the system is working properly. We also record the ankle angle signal from the encoder.

Fig.4.3 DAQ card USB-6251

4.3. Experimental result

4.3.1 Level walking result

This experiment is tested on a treadmill and walking speed at 2km/h. Data collection was from DAQ IN-6521. The voltage on the X-axis and Y-axis is sampling data, sampling rate is 2000/s. The data collection (Fig.4.4) of 4 steps walking data, the (Fig.4.4 (a)) is a healthy walk signal from A1, (Fig.4.4 (b)) is an artificial signal from A2, the walking start in swing phase when heel-strike the heel just touches the ground until the top touches the ground, and there had a great change in Z axis acceleration. During the stance phase Z axis acceleration almost unchanged. During the swing phase when the top is off the ground, the top in the Z axis has a larger swing and it is be reflected in the acceleration. From this result healthy and artificial ankle has similar phase change, but it also have noise in the second time swing phase, this noise is from elastic shock in swing phase.

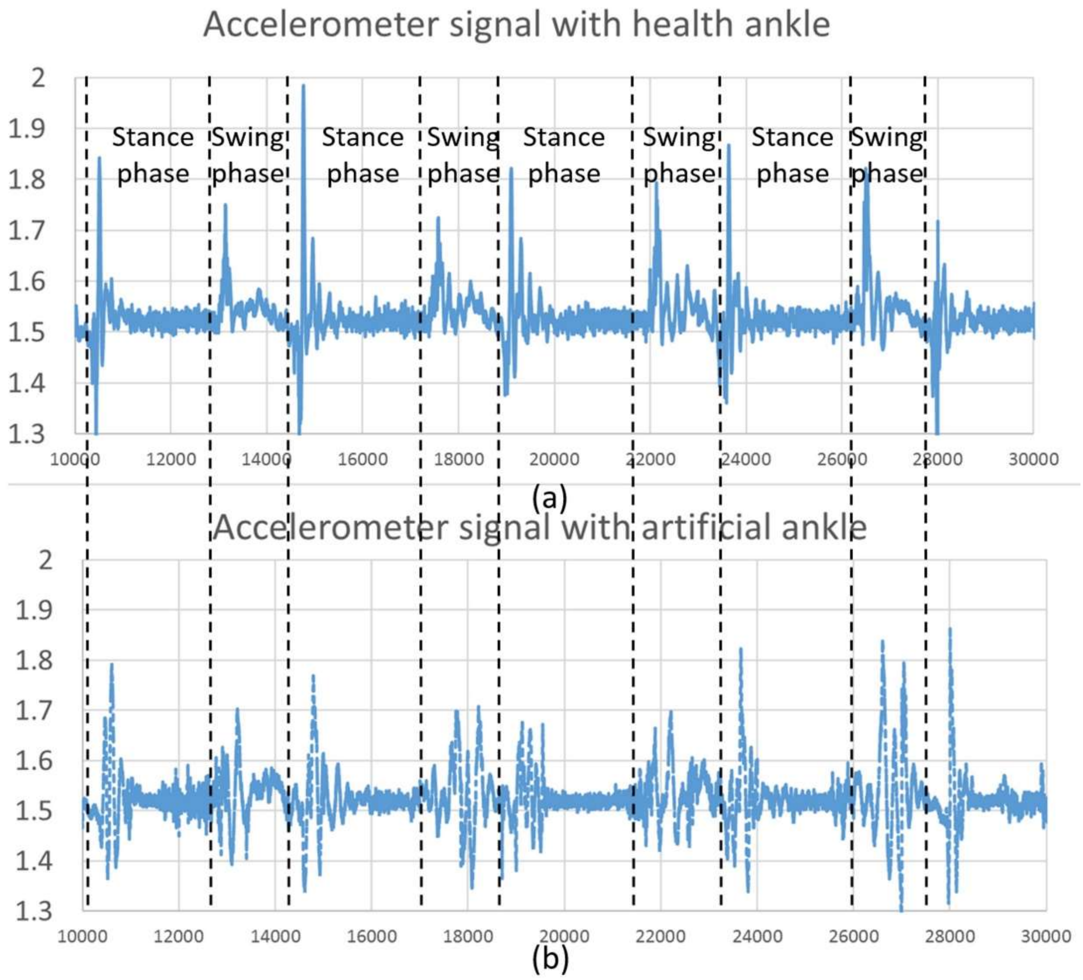


Fig.4.4 level-walking result at a speed of 2km/h sampling rate is 2000/s, a: health ankle b: artificial ankle

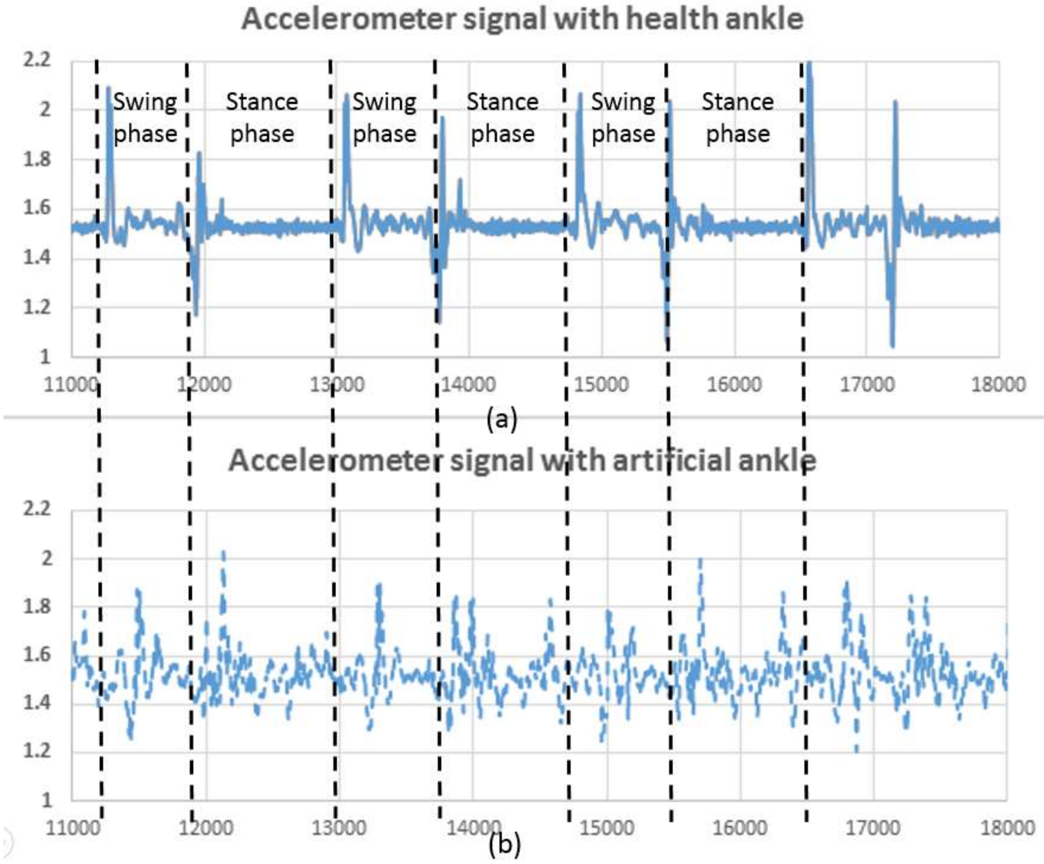


Fig.4.5 level-walking result at a speed of 3km/h sampling rate is 2000/s, a: health ankle b: artificial ankle

For the walking speed at 3km/h (Fig.4.5 4.7), the encoder in accordance with the reference data (Fig.4.7). But the signal from accelerometer, greater speed will lead to more shock, shock causes increased noise in stance phase. This is the limitation for this artificial model.

The ankle angle signal from the encoder and compare with reference data, chart (Fig.4.6 (a)) is reference data from [15], and second chart is artificial ankle data (Fig.4.6 (b)). Walking phase is described as follows:

1. Heel-strike to foot-flat: 0 to 20 percent, the heel initially make contact with the ground and the top was follow. In this phase ankle angle from 0^{o} to 10^{o}.

2. Foot-flat to max dorsiflexion: 20 to 60 percent, begins at the foot-flat and continues until the ankle maximum angle ankle angle from 10^{o} to -7^{o}.

3. Swing phase: 60 to 90 percent, the foot off the ground and ankle angle reset to 0^{o}.

Comparison with reference data [15]. The changing of walking phase is similar, the difference is slope during heel-strike to foot-flat, the problem is from the DC-motor speed was faster than human, this problem can be solve in a mechanical system such as gear transmission system. This system can effectively control the ankle angle in each phase during the level-walking.

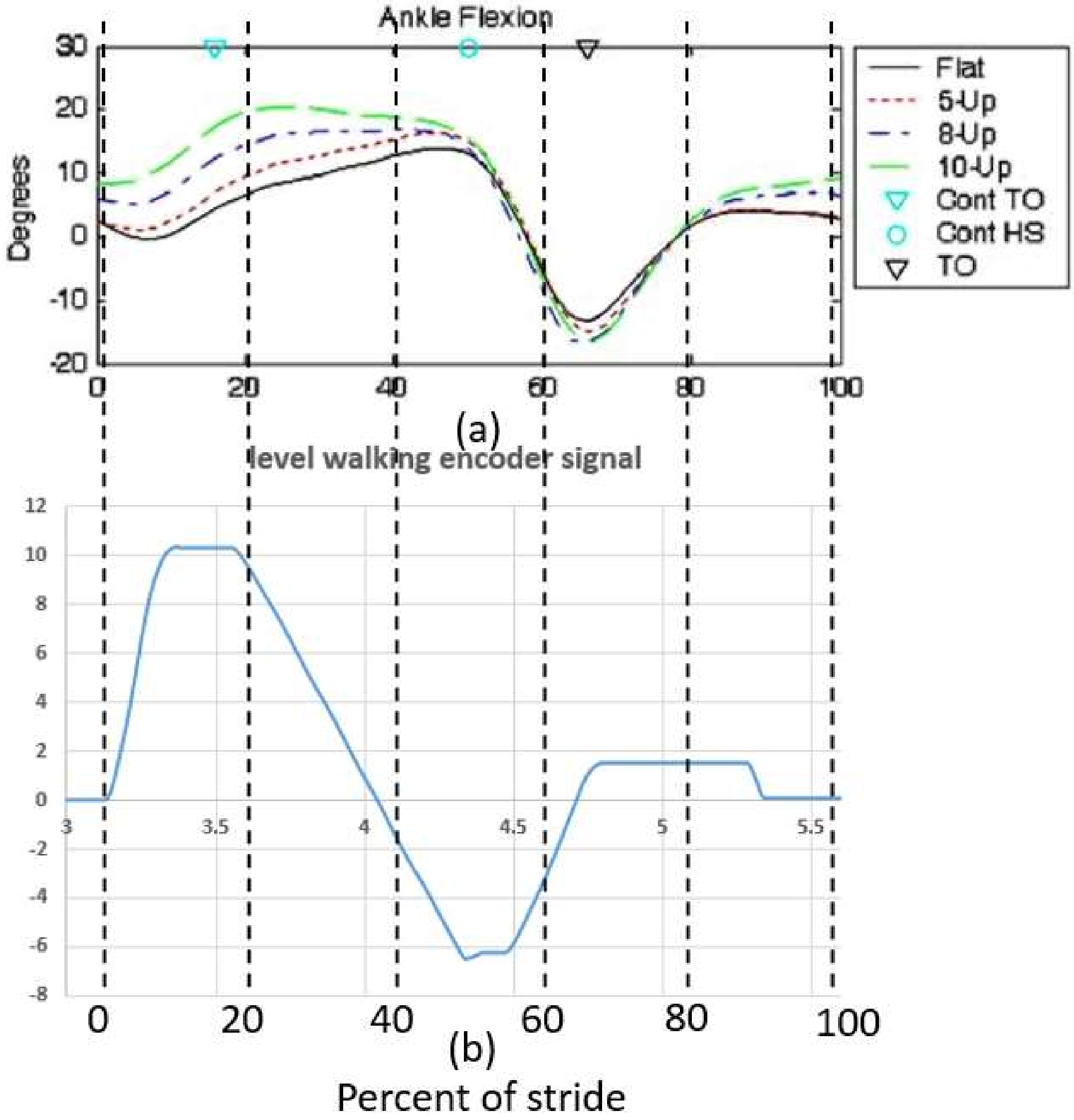


Fig.4.6 Ankle angle data from level walking at a speed of 2km/h. (a): health walking (b): artificial ankle degree

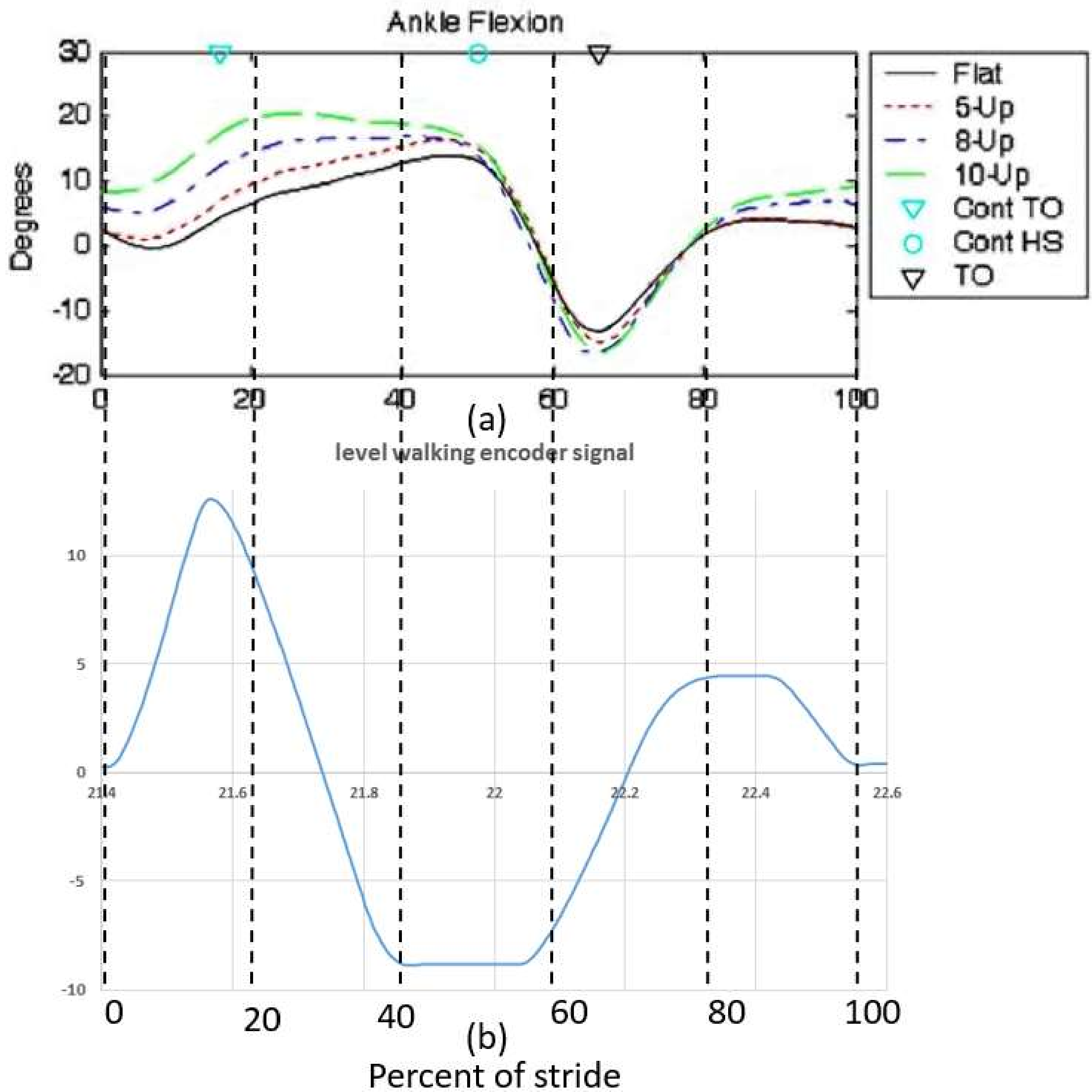


Fig.4.7 Ankle angle data from level walking at a speed of 3km/h. (a): health walking (b): artificial ankle degree

Fig.4.7 was the encoder signal form level walking at a speed of 3km/h, from this result between 70-80 percent, the ankle degree have error about 5°, this error is from the motor inertia and the system will correct the errors in 90 percent gait cycle.

4.3.2 10° upslope walking result

From biomechanics of normal gait research [15], the relationship between upslope walking and level-walking is the change of walking phase was be similar, but the ankle degree in each gait phase was different (Fig.2.1). Compare with level-walking, the 10° upslope walking, the start of ankle degree was change to 10° and heel-strike max ankle degree was change to 20°.Reset the reset the artificial ankle model parameter to 10°

upslope walking, working result as follows (Fig.4.8 4.9(b)):

1. Heel-strike to foot-flat: 0 to 20 percent, the heel initially make contact with the ground and the top was follow .in this phase ankle angle from 10° to 20°.

2. Foot-flat to max dorsiflexion: 20 to 60 percent, begins at the foot-flat and continues until the ankle maximum angle ankle angle from 20° to -10°.

3. Swing phase: 60 to 90 percent, the foot off the ground and ankle angle reset to 10°.

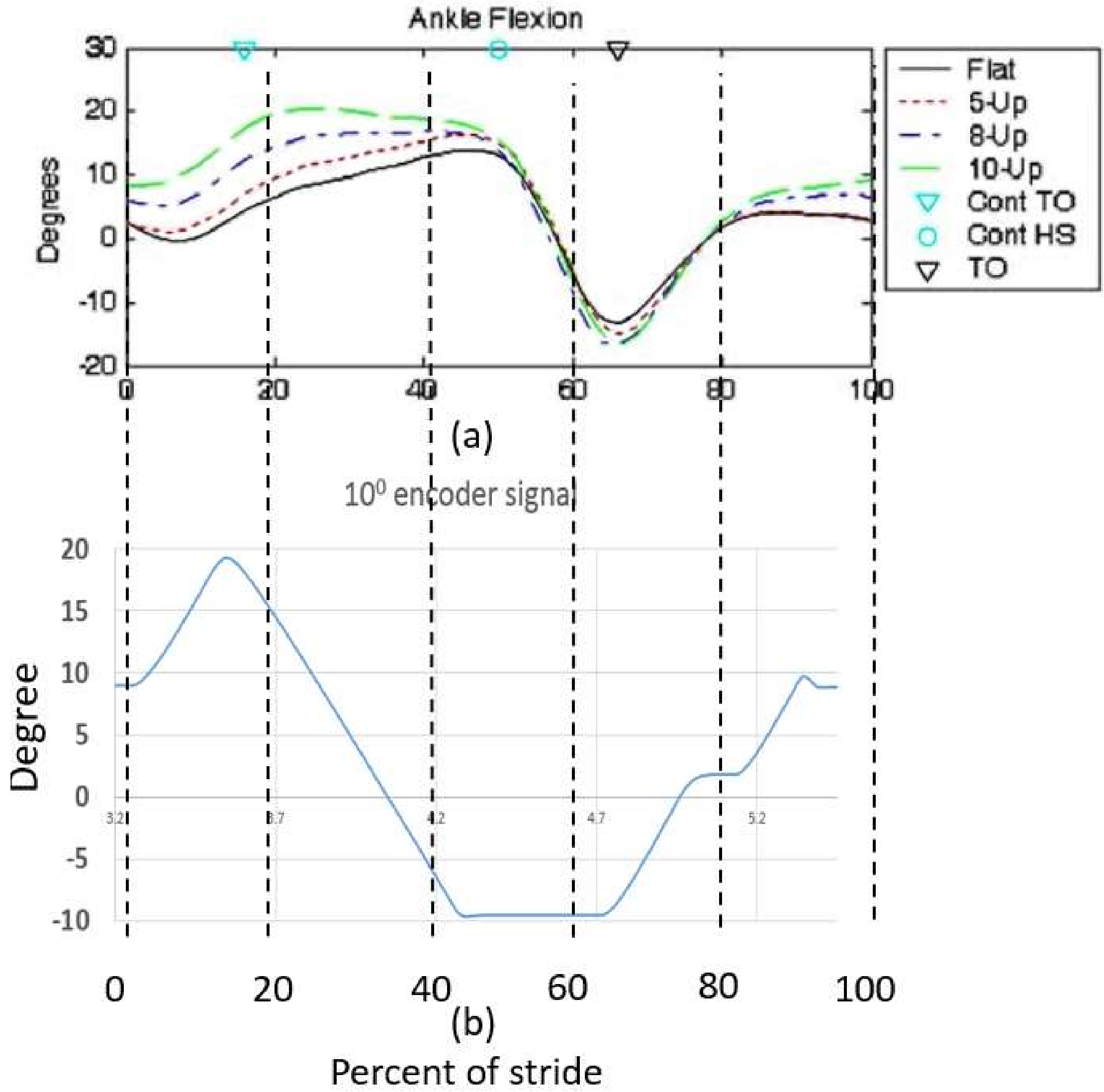


Fig.4.8 Ankle angle data from 10° upslope walking at a speed of 2km/h. (a): health walking (b): artificial ankle degree

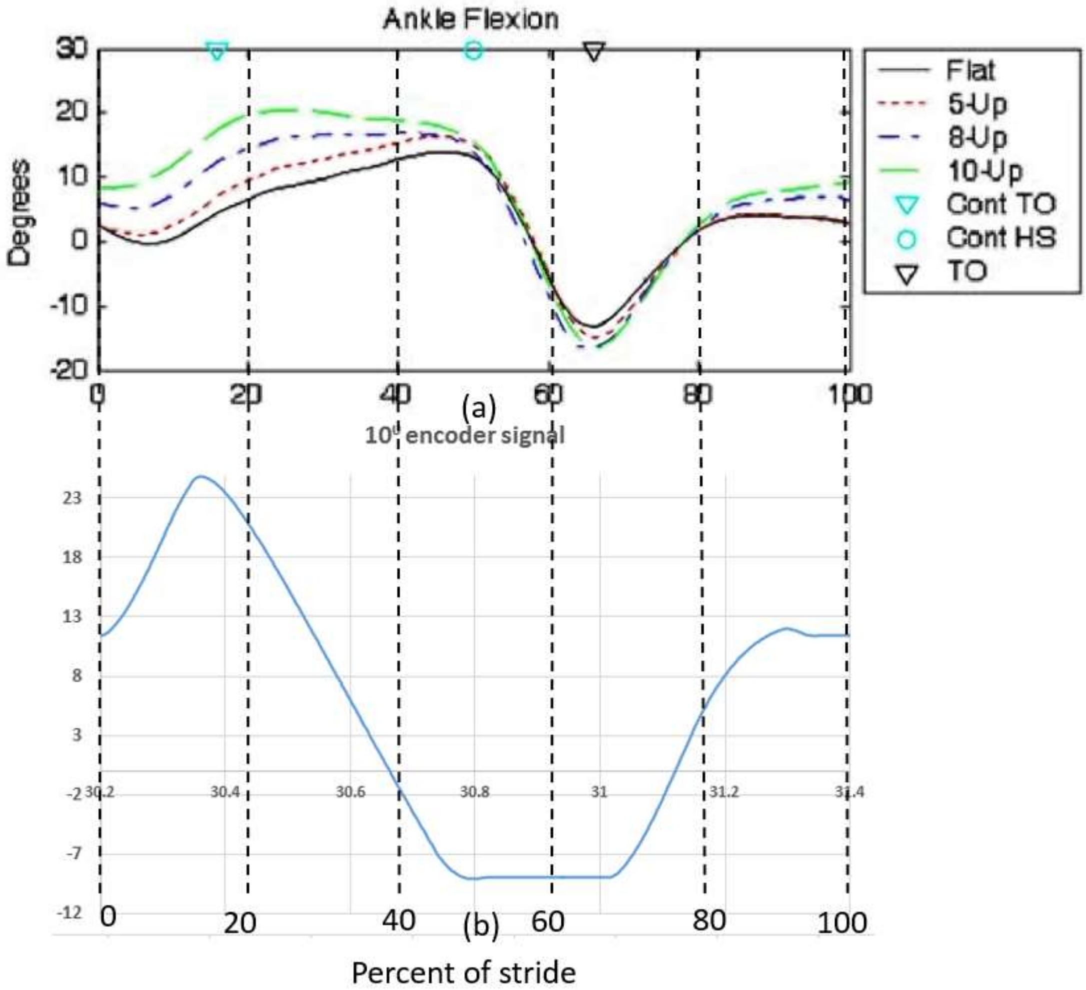


Fig.4.9 Ankle angle data from 10° upslope walking at a speed of 3km/h. (a): health walking (b): artificial ankle degree

In order to check the working state of system, record the accelerometer signal and compare. This experiment is tested on a treadmill and walking speed is 2km/h. The data from DAQ IN-6521 sampling rate is 1000/s. The data collection (Fig.4.10) of 3 steps walking data, healthy walk signal from A1 in (Fig.4.10 (a)). The (Fig.4.10 (b)) is artificial signal from A2. For the comparison, at the beginning and end of each phase, change of Z-axis acceleration from healthy and artificial was similar. Between the stance phase sole of foot touch the ground, Z-axis acceleration in this phase was zero, but in (Fig.4.10(b)) the signal between the stance phase still have change , artificial model has swing in this phase, that swing from Plastic elastic vibration. But for walking speed at 3km/h, it has same problem with 3km/h level walking speed, this artificial model have limitation in 3km/h walking speed. In summary this system can working stably in 10° upslope walking at 2km/h walking speed.

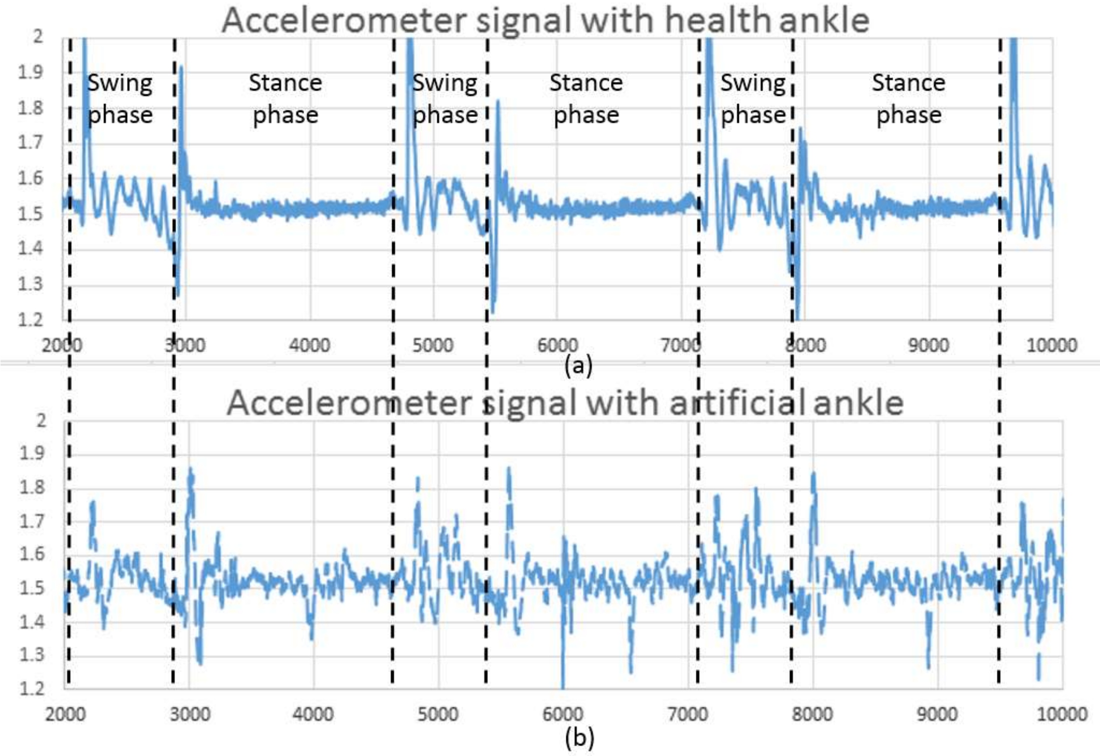


Fig.4.10 10^{o} upslope walking result at a speed of 2km/h sampling rate is 1000/s, a: health ankle b: artificial ankle

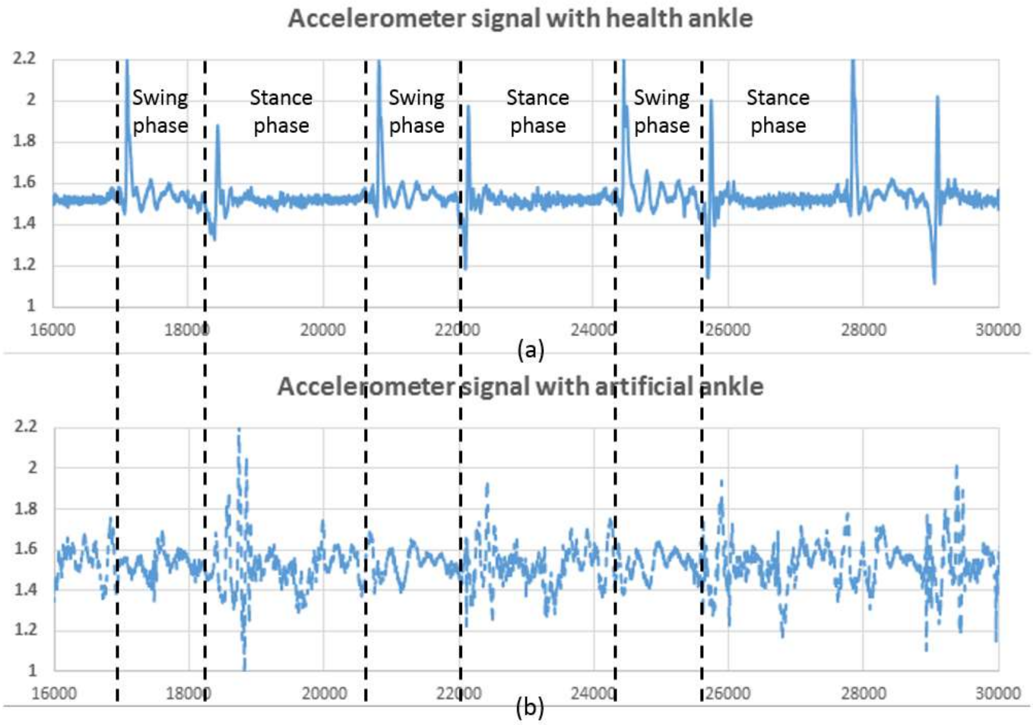


Fig.4.11 10^o upslope walking result at a speed of 3km/h sampling rate is 1000/s, a: health ankle b: artificial ankle

4.3.3 5^0 upslope walking result

In 5^0 upslope walking the ankle degree start in 0^o but in heel-strike phase the max ankle degree change from10^o to 15^o (Fig.4.12 (a)). Therefore reset the artificial ankle model parameter to 5^o upslope walking data: the red line in (Fig.4.12 (a)). And record the working status as follows (Fig.4.12 (b)):

1. Heel-strike to foot-flat: 0 to 20 percent, the heel initially make contact with the ground and the top was follow .in this phase ankle angle from 0^o to 15^o.

2. Foot-flat to max dorsiflexion: 20 to 60 percent, begins at the foot-flat and continues until the maximum ankle angle from 15^o to -10^o.

3. Swing phase: 60 to 90 percent, the foot off the ground and ankle angle reset to 0^o.

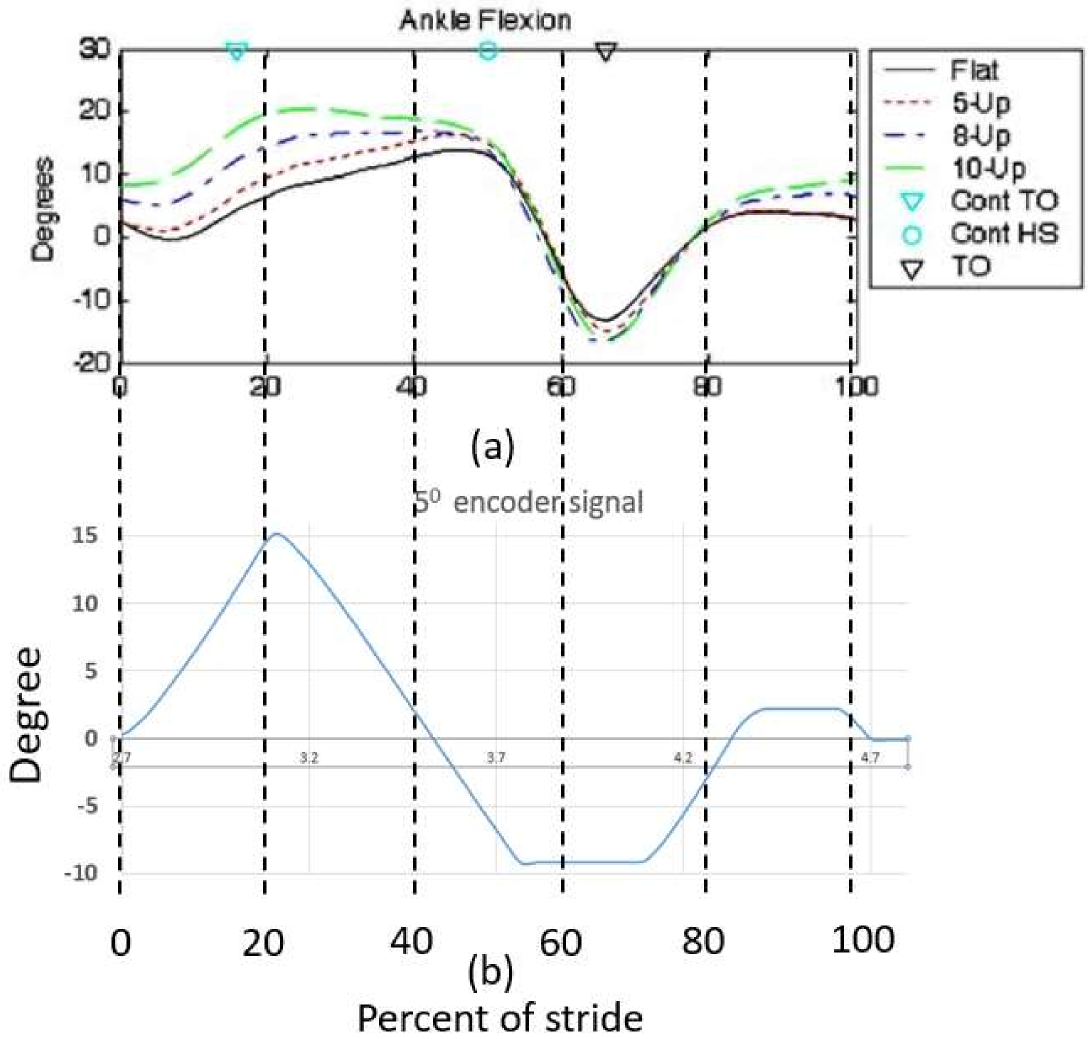


Fig.4.12 Ankle angle data from 5^0 upslope walking. (a):health walking (b): artificial ankle degree

This experiment is tested on a treadmill and walking speed is 1.5km/h. The data from DAQ IN-6521 sampling rate is 1000/s. The data collection (Fig.4.13) of 2 steps walking data, the (Fig.4.13 (a)) is a healthy walk signal from A1, (Fig.4.13 (b)) is an artificial signal from A2. In this result the walking phase change can also be reflected in accelerometer Z axis, the acceleration have a great change between swing phase and next step stance phase, but during the stance phase the signal from artificial ankle have a lot of noise, this noise is from plastic elastic vibration, The results are worse than expected.

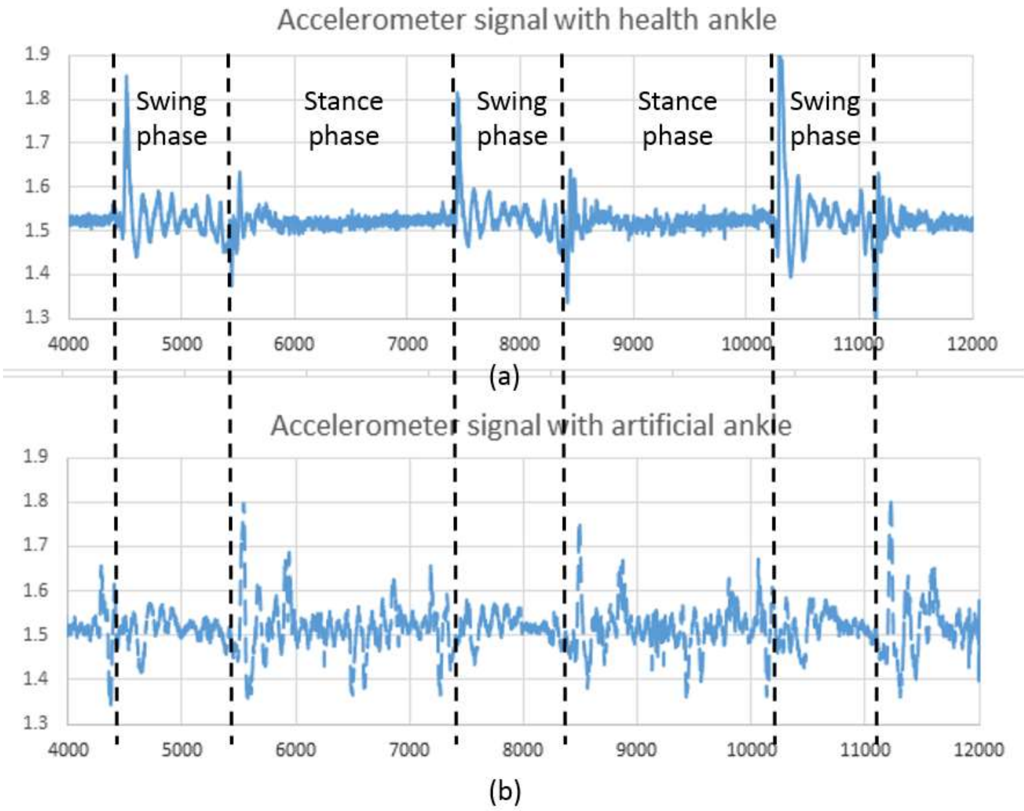


Fig.4.13 5° upslope walking result at a speed of 1.5km/h, sampling rate is 1000/s, a: health ankle b: artificial ankle

In 5^0 2km/h upslope walking, the signal have big noise from accelerometer in swing and stance phase. Because the accelerometer cannot measure the fast walking when the plastic model has large shock. But the encoder signal can show the model working was correct (Fig.4.12). This artificial model limitation at 5° upslope is 2km/h walking speed.

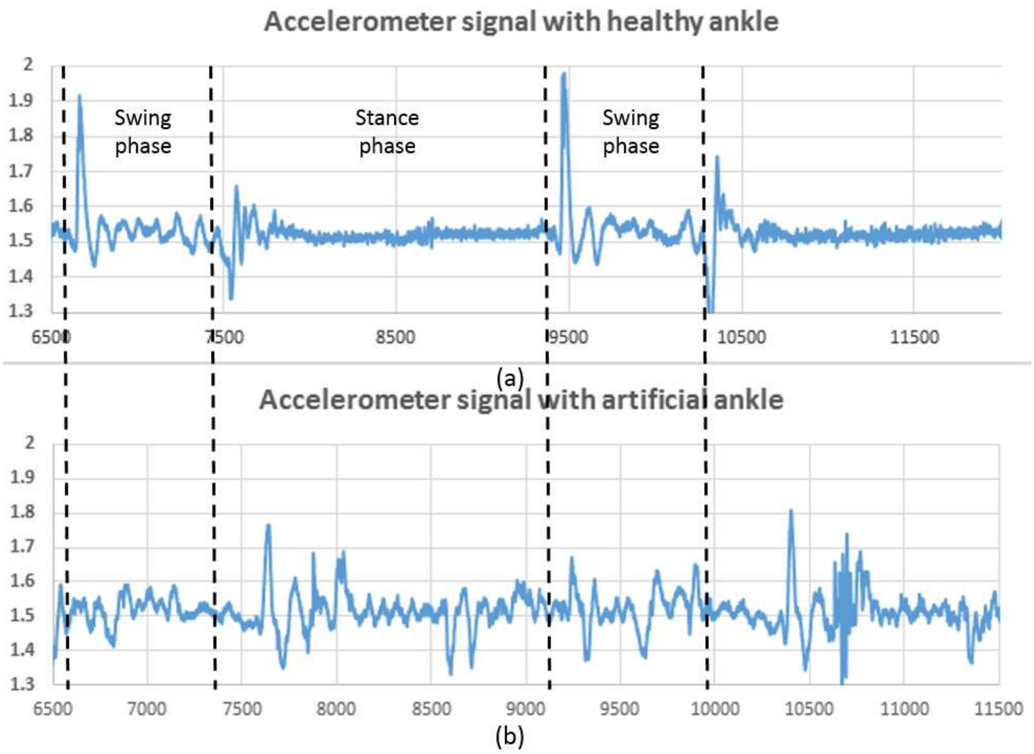


Fig.4.14 5° upslope walking result at a speed of 2km/h sampling rate is 1000/s, a: health ankle b: artificial ankle

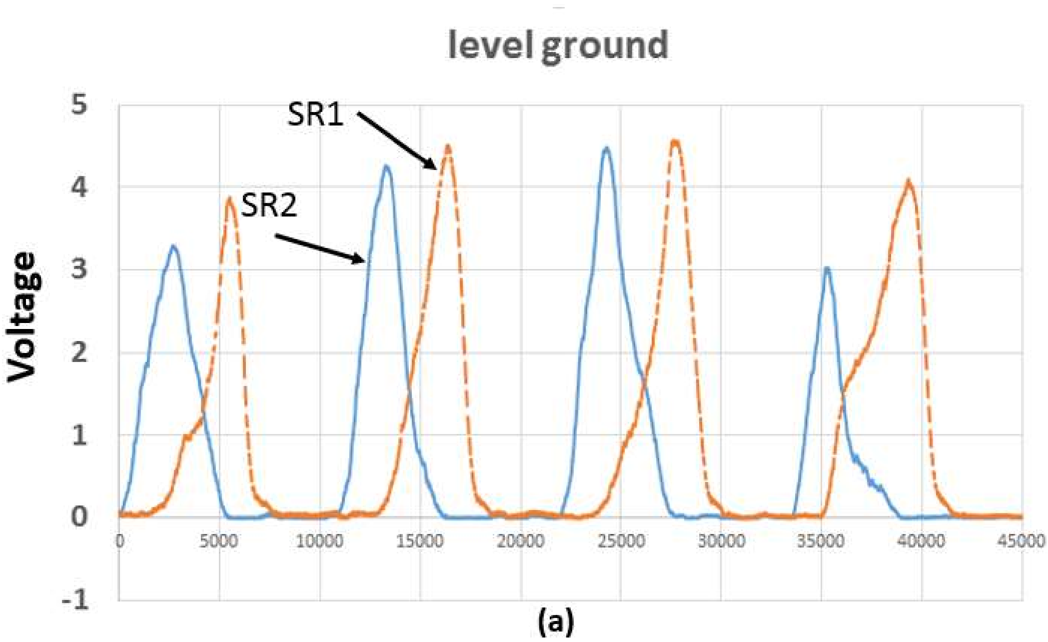
4.3.4 Flexi-force sensor signal with level and 10° upslope walking

The Flexi-force sensor not suitable for used in the system switch from level-ground to upslope walking. The sensor output signal will follow V_{out} = Body weight + F_{leg}, F_{leg} is the force from the muscles. The body weight is fix but the F_{leg} will change in every step. The V_{out} was different with every step. The walking signal from level-ground and 10° upslope walking as follow (Fig.4.15):

1. Level-ground: SR2 maximum output in each step was from 3V-4.5V, SR1 maximum output in each step was from 4V-4.5V.

2. 10° upslope: SR2 maximum output in each step was from 2.2V-3V, SR1 maximum output in each step was from 4V- 4.5V.

To sum up they have same output in different walking states. The system using Flexi-force sensor was difficult to distinguish level-ground from upslope walking.



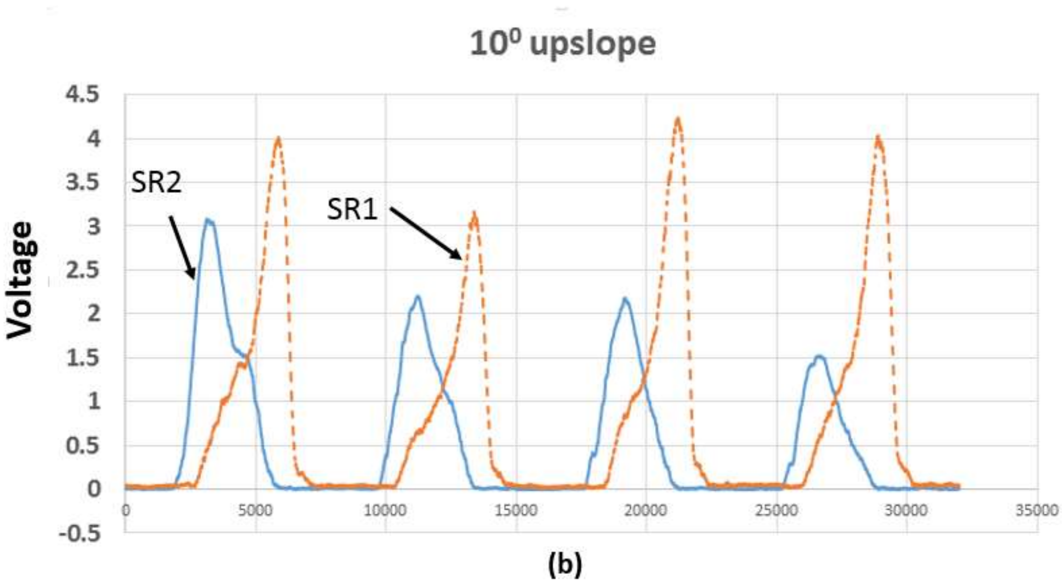


Fig.4.15 walking signal from level and upslope sampling rate is 4000/s (a): level ground (b):10^o upslope line: SR2 Dash line: SR1

4.3.5 System working with load and no-load

It is worth mentioning that all experiments were tested in no-load walking, the system peak current with no-load working was 700mA. We also install the 1kg weight under the power ankle model (Fig.4.16).

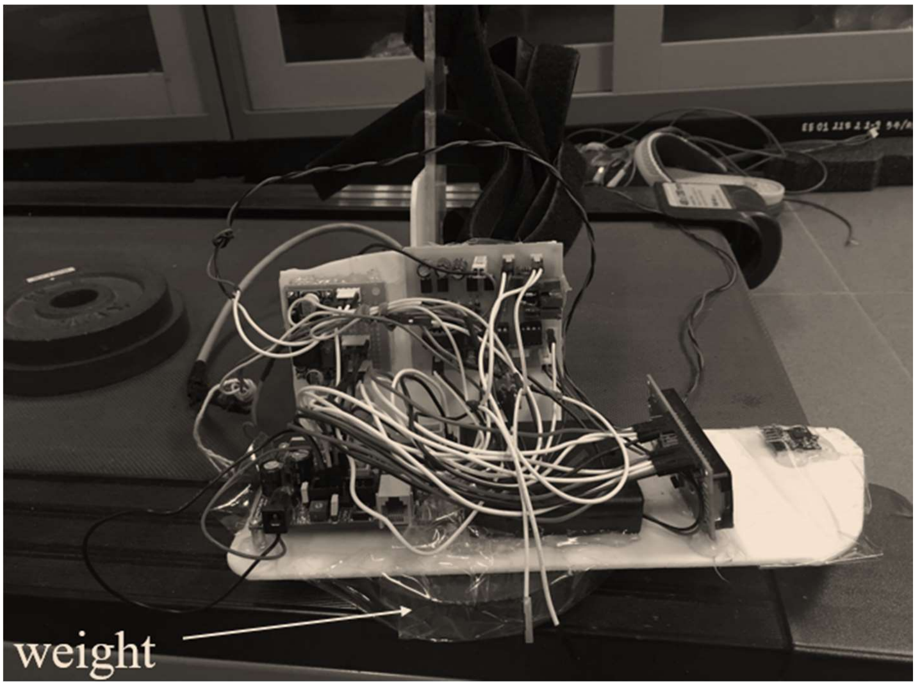


Fig.4.16 Power ankle model weight position.

Before the experiment system working data was set for level walking and 3km/h walking speed, we recording and compare the encoder signal in no-load and load working. The walking signal from no-load and load walking as follow (Fig.4.17):

No-load: one gait need 1.2 second (Fig.4.17 (a)), peak current was 700mA.

Load: one gait need 1.2 second (Fig.4.17 (b)), peak current was 800mA.

From the result we can know the system response time with no-load and load ware similar. And in the load working peak current higher than no-load, the peak current depend on walking load.

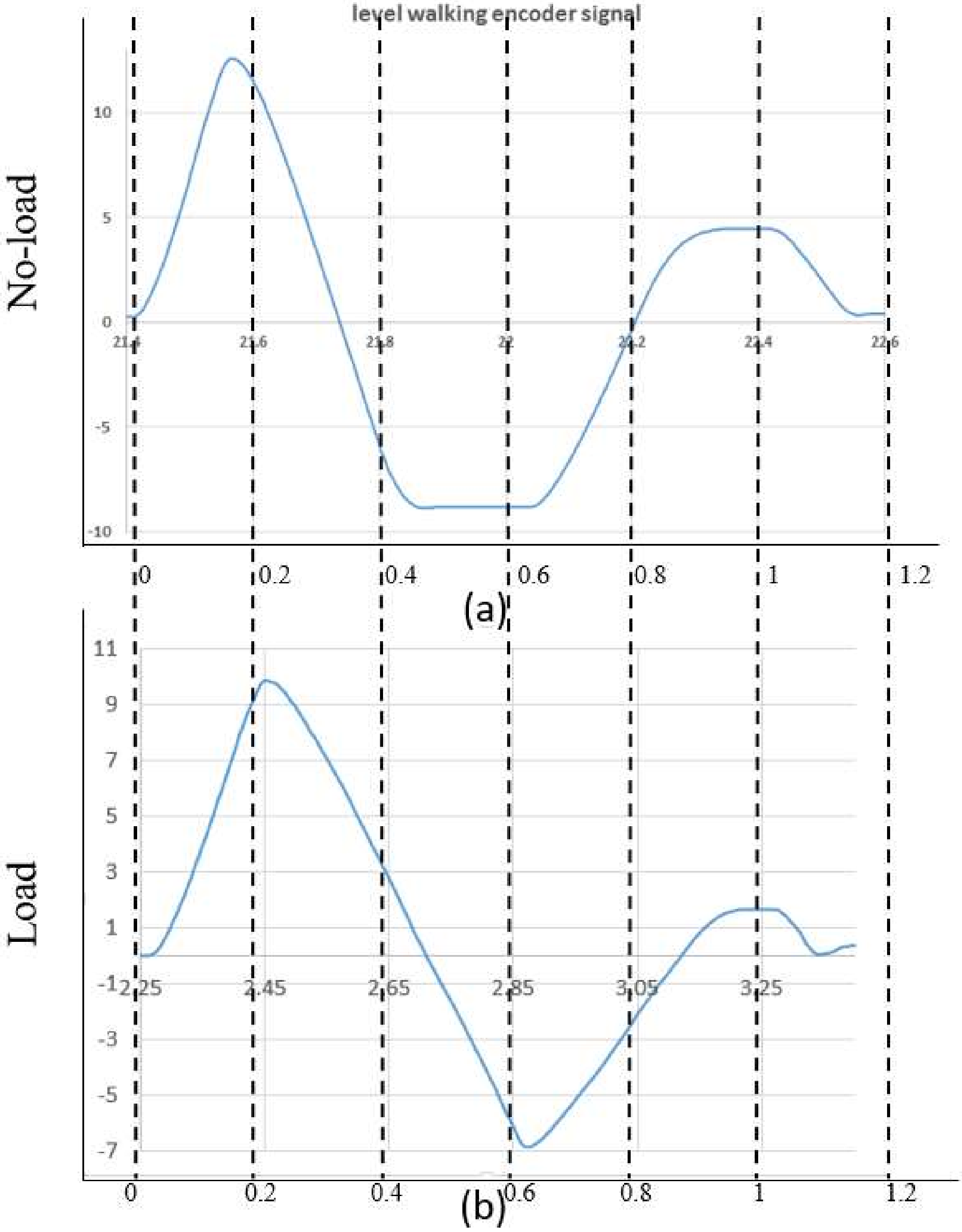


Fig.4.17 Encoder signal for level walking with no-load and load (a): no-load (b): load x-axis: time Y-axis: degree

Chapter 5
Discussion and Conclusion

5.1 Discussion

The traditional control methods for power ankle [3]-[10] are consists of two groups: high complex calculation from inertial measurement unit or complex mechanical system. The inertial measurement unit need high performs computation system it would use more energy, the latter complex mechanical system difficult to maintain. The control system based on Flexi-force sensor as an on-off variable, system has the following advantage of simple calculation, fast system response time and low energy consumption.

The limited experimental conditions, the control system only tested on the artificial model and the Flexi-force sensor installed under the sole with healthy foot, and compare the acceleration between the healthy foot and artificial model, the experimental process is similar gait copy. This artificial mode limitation as follow:

1. Level-ground walking: The system can working in 2km/h, limitation is 3km/h.
2. 10^{o} upslope walking: The system can working in 2km/h, limitation is 3km/h.
3. 5^{o} upslope walking: The system working in 1.5km/h were worse than expected, limitation is 2km/h.

This problem is from the artificial mode, plastic has large shock during walking, the more faster walking speed more large shock. This test result did not give any suggestion that it can work well with the artificial power ankle. Therefore we must installed the Flexi-force sensor to real artificial power ankle, and test this system whether it can work independently without healthy foot.

During the walking, the Flexi-force sensor was used as an on-off variable， control the ankle at the beginning of the phase, the ankle stop depend on ankle position feedback from encoder. Therefore the ankle speed must be setting before walking, the speed was fix during the walking. In the Chapter3.3 walking gait phase measurement we also find the slope of walking signal have relationship with walking speed, but it just a hypothesis, it had more experiment to find that relation equation, use to change the walking speed in real-time.

This system can be used in level, 5^{o} and 10^{o} upslope walking, but it was discontinuity, the walking parameter had to be settled before walking environment change, this problem is from slope estimate, the system continuity work must estimate the slope before upslope, but the Flexi-force sensor only can detect walking in real-time, cannot estimate. For this problem we have two way to solve it. First is to use other sensor to detect the upslope before walking, such as laser. The other way is install one more Flexi-force sensor under the healthy foot for reference data, when upslope healthy foot is first step, the Flexi-force sensor signal was different between the different upslope. Therefore it also have more experiment to perfect this method.

5.2. Conclusion

This thesis consists of two part: walking phase measurement and artificial power ankle control system design.

Flexi-force sensor is a good solution at the walking phase walking phase measurement. It is so easy to analysis data using two Flexi-force sensor. The sensor SR1 installed under the fore foot and sensor SR2 installed under the heel. It can accurately measure each walking phase during the walking in real-time (Fig.2.1). The conventional walking phase measurement using inertial measurement unit must use complex algorithm. Flexi-force sensor SR1 and SR2 walking phase measurement only compare the two sensor result as flows:

1. Swing phase: SR1< threshold and SR2< threshold.

2. Heel-strike to max dorsiflexion phase: SR2> threshold and SR1 < threshold.

3. Max dorsiflexion phase to swing phase: SR2< threshold and SR1 > threshold.

The control system based on the Flexi-force sensor can be straightforward measurement the walking phase during the real-time walking. Experiment results showed that is a great way for power ankle control system. It is more straightforward than traditional control methods impedance control or inertial sensor.

The system testing in level-ground, 5^{o}, 10^{o} upslope walking. The walking result as follows:

1. Level-ground: system can Stability work in 2km/h walking speed. The artificial power ankle phase transformation is similar to healthy ankle.

2. 10^{o} upslope walking : system in 10^{o} upslope also can Stability working in 2km/h walking speed, but the result shows it have noise in stance phase, this

noise was from plastic elastic vibration.

3. 5^0 upslope walking: Although the system can working in 1.5km/h walking speed but the result were worse than expected.

4. 15^0 upslope walking: In 15^0 upslope walking the walking phase was different with normal walking because of the slope was too high, The test found that ankle degree in 15^0 upslope always keep to the same degree, no heel-strike no foot-flat only Max-dorsiflexion. So 15^0 upslope walking was meaningless.

For system power, this system power by 12V DC, system standby current was 200mA, working peak current no more than 700mA, so the system power no more than 8W, the 12V 8000mah battery group can provide the 8W power for 12 hour ,so it can working about 12 hour.

5.3. Recommendations for future work

Firstly, as the limited experimental conditions. The future works are suggest. All of experiment was test on the plastic artificial model. However, the experiment did not give any suggestion that it can work on real artificial power ankle. Therefore, to make a real artificial power ankle and testing this system are necessary. To get this system defect and improve it to perfect the system.

Secondly, although this system can working in level-ground, 5^o, 10^o upslope walking but the system parameter had to reset before walking, is non-adaptive. Therefore, to develop more control method are necessary.

Flexi-force sensor can be develop more movement in artificial power ankle control such as: downhill slopes, Up and down stairs.

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Appendix A Flexi-force sensor

FlexiForce®
A201 Standard Force & Load Sensors

Actual size of sensor

Physical Properties

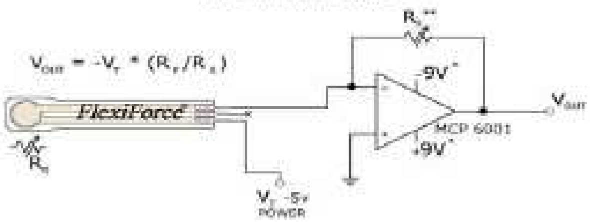
Thickness	0.008" (0.208 mm)
Length	7.75" (197 mm), *optional trimmed lengths*: 6" (152 mm), 4" (102 mm), or 2" (51mm)
Width	0.55" (14 mm)
Sensing Area	0.375" diameter (9.53 mm)
Connector	3-pin Male Square Pin (center pin is inactive)
Substrate	Polyester (ex: Mylar)

Standard Force Ranges (as tested with circuit shown below)

0 - 1 lb. (4.4 N)
0 - 25 lb. (110 N)
0 - 100 lb. (440 N)*

In order to measure forces above 100 lb (up to 1000 lb), apply a lower drive voltage and reduce the resistance of the feedback resistor (1kΩ min.)

Recommended Circuit



- * Supply Voltages should be constant
- ** Reference Resistance R_F is 1kΩ to 100kΩ
- Sensor Resistance R_S at no load is >5MΩ
- Max recommended current is 2.5mA

Typical Performance		Evaluation Conditions
Linearity (Error)	±3%	Line drawn from 0 to 50% load
Repeatability	±2.5% of full scale	Conditioned sensor, 80% of full force applied
Hysteresis	< 4.5 % of full scale	Conditioned sensor, 80% of full force applied
Drift	< 5% per logarithmic time scale	Constant load of 25 lb (111 N)
Response Time	< 5 μsec	Impact load, output recorded on oscilloscope *Time required for the sensor to respond to an input force*
Operating Temperature	15°F - 140°F (-9°C - 60°C)*	
Output Change/Degree F	±0.2%/°F (0.36%/°C)	

*For loads less than 10 lbs, the operating temperature can be increased to 165°F (74°C)

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Rev G_021109

Tekscan

Appendix B DSPIC30F2010

dsPIC30F2010

High-Performance, 16-bit Digital Signal Controller

Note:	This data sheet summarizes features of this group of dsPIC30F devices and is not intended to be a complete reference source. For more information on the CPU, peripherals, register descriptions and general device functionality, refer to the "*dsPIC30F Family Reference Manual*" (DS70046). For more information on the device instruction set and programming, refer to the "*16-bit MCU and DSC Programmer's Reference Manual*" (DS70157).

High-Performance Modified RISC CPU:

- Modified Harvard architecture
- C compiler optimized instruction set architecture
- 83 base instructions with flexible addressing modes
- 24-bit wide instructions, 16-bit wide data path
- 12 Kbytes on-chip Flash program space
- 512 bytes on-chip data RAM
- 1 Kbyte nonvolatile data EEPROM
- 16 x 16-bit working register array
- Up to 30 MIPs operation:
  - DC to 40 MHz external clock input
  - 4 MHz-10 MHz oscillator input with PLL active (4x, 8x, 16x)
- 27 interrupt sources
- Three external interrupt sources
- Eight user-selectable priority levels for each interrupt
- Four processor exceptions and software traps

DSP Engine Features:

- Modulo and Bit-Reversed modes
- Two 40-bit wide accumulators with optional saturation logic
- 17-bit x 17-bit single-cycle hardware fractional/integer multiplier
- Single-cycle Multiply-Accumulate (MAC) operation
- 40-stage Barrel Shifter
- Dual data fetch

Peripheral Features:

- High current sink/source I/O pins: 25 mA/25 mA
- Three 16-bit timers/counters; optionally pair up 16-bit timers into 32-bit timer modules
- Four 16-bit capture input functions
- Two 16-bit compare/PWM output functions
  - Dual Compare mode available
- 3-wire SPI modules (supports 4 Frame modes)
- I²C™ module supports Multi-Master/Slave mode and 7-bit/10-bit addressing
- Addressable UART modules with FIFO buffers

Motor Control PWM Module Features:

- Six PWM output channels
  - Complementary or Independent Output modes
  - Edge and Center-Aligned modes
- Four duty cycle generators
- Dedicated time base with four modes
- Programmable output polarity
- Dead-time control for Complementary mode
- Manual output control
- Trigger for synchronized A/D conversions

Quadrature Encoder Interface Module Features:

- Phase A, Phase B and Index Pulse input
- 16-bit up/down position counter
- Count direction status
- Position Measurement (x2 and x4) mode
- Programmable digital noise filters on inputs
- Alternate 16-bit Timer/Counter mode
- Interrupt on position counter rollover/underflow

Analog Features:

- 10-bit Analog-to-Digital Converter (ADC) with:
  - 1 Msps (for 10-bit A/D) conversion rate
  - Six input channels
  - Conversion available during Sleep and Idle
- Programmable Brown-out Reset

dsPIC30F2010

Special Digital Signal Controller Features:

- Enhanced Flash program memory:
  - 10,000 erase/write cycle (min.) for industrial temperature range, 100K (typical)
- Data EEPROM memory:
  - 100,000 erase/write cycle (min.) for industrial temperature range, 1M (typical)
- Self-reprogrammable under software control
- Power-on Reset (POR), Power-up Timer (PWRT) and Oscillator Start-up Timer (OST)
- Flexible Watchdog Timer (WDT) with on-chip low-power RC oscillator for reliable operation
- Fail-Safe Clock Monitor (FSCM) operation
- Detects clock failure and switches to on-chip Low-Power RC (LPRC) oscillator
- Programmable code protection
- In-Circuit Serial Programming™ (ICSP™) programming capability
- Selectable Power Management modes
  - Sleep, Idle and Alternate Clock modes

CMOS Technology:

- Low-power, high-speed Flash technology
- Wide operating voltage range (2.5V to 5.5V)
- Industrial and Extended temperature ranges
- Low power consumption

dsPIC30F Motor Control and Power Conversion Family

Device	Pins	Program Mem. Bytes/ Instructions	SRAM Bytes	EEPROM Bytes	Timer 16-bit	Input Cap	Output Comp/Std PWM	Motor Control PWM	A/D 10-bit 1 Msps	QEI	UART	SPI	I²C™
dsPIC30F2010	28	12K/4K	512	1024	3	4	2	6 ch	6 ch	Yes	1	1	1

Appendix C IR2110

International
IOR Rectifier

Data Sheet No. PD60147 rev.U

IR2110(-1-2)(S)PbF/IR2113(-1-2)(S)PbF

HIGH AND LOW SIDE DRIVER

Features

- Floating channel designed for bootstrap operation
  Fully operational to +500V or +600V
  Tolerant to negative transient voltage
  dV/dt immune
- Gate drive supply range from 10 to 20V
- Undervoltage lockout for both channels
- 3.3V logic compatible
  Separate logic supply range from 3.3V to 20V
  Logic and power ground ±5V offset
- CMOS Schmitt-triggered inputs with pull-down
- Cycle by cycle edge-triggered shutdown logic
- Matched propagation delay for both channels
- Outputs in phase with inputs

Product Summary

V_{OFFSET} (IR2110)	500V max.
(IR2113)	600V max.
I_O+/-	2A / 2A
V_{OUT}	10 - 20V
$t_{on/off}$ (typ.)	120 & 94 ns
Delay Matching (IR2110)	10 ns max.
(IR2113)	20ns max.

Packages

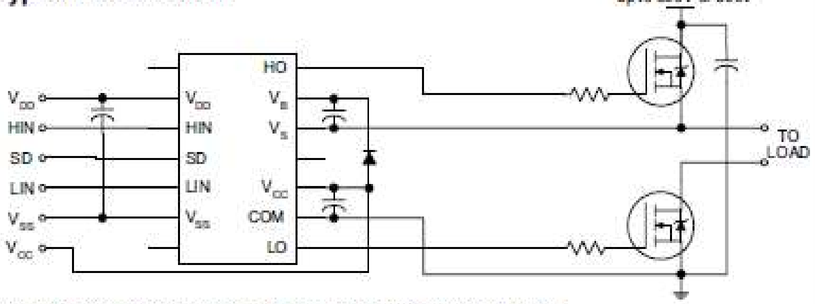
14-Lead PDIP
IR2110/IR2113

16-Lead SOIC
IR2110S/IR2113S

Description

The IR2110/IR2113 are high voltage, high speed power MOSFET and IGBT drivers with independent high and low side referenced output channels. Proprietary HVIC and latch immune CMOS technologies enable ruggedized monolithic construction. Logic inputs are compatible with standard CMOS or LSTTL output, down to 3.3V logic. The output drivers feature a high pulse current buffer stage designed for minimum driver cross-conduction. Propagation delays are matched to simplify use in high frequency applications. The floating channel can be used to drive an N-channel power MOSFET or IGBT in the high side configuration which operates up to 500 or 600 volts.

Typical Connection



(Refer to Lead Assignments for correct pin configuration). This/These diagram(s) show electrical connections only. Please refer to our Application Notes and DesignTips for proper circuit board layout.

Appendix D IRFD220

intersil

IRFD220

Data Sheet | July 1999 | File Number 2317.3

0.8A, 200V, 0.800 Ohm, N-Channel Power MOSFET

This N-Channel enhancement mode silicon gate power field effect transistor is an advanced power MOSFET designed, tested, and guaranteed to withstand a specified level of energy in the breakdown avalanche mode of operation. All of these power MOSFETs are designed for applications such as switching regulators, switching convertors, motor drivers, relay drivers, and drivers for high power bipolar switching transistors requiring high speed and low gate drive power. These types can be operated directly from integrated circuits.

Formerly developmental type TA09600.

Ordering Information

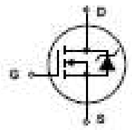
PART NUMBER	PACKAGE	BRAND
IRFD220	HEXDIP	IRFD220

NOTE: When ordering, use the entire part number.

Features

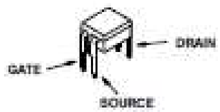
- 0.8A, 200V
- $r_{DS(ON)} = 0.800\Omega$
- Single Pulse Avalanche Energy Rated
- SOA is Power Dissipation Limited
- Nanosecond Switching Speeds
- Linear Transfer Characteristics
- High Input Impedance
- Related Literature
  - TB334 "Guidelines for Soldering Surface Mount Components to PC Boards"

Symbol



Packaging

HEXDIP



Appendix E Encoder

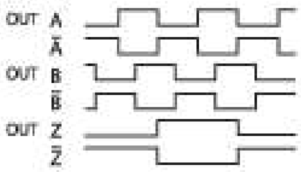
PR-01/PR-02

INCREMENT ENCODER

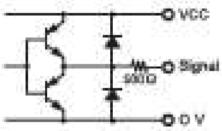
DESCRIPTION

- High Accuracy and wide applications
- Size : 50 mm. Shaft : 8 mm.
- Power Supply : 5-30 VDC
- Push-Pull output type. Can replace NPN and PNP open collector
- Variety sizes PR are available.

WAVE FROM



OUTPUT DIAGRAM



WIRING DIAGRAM

		PR-01 Max Speed	
Brown	+8.0t o3 0.0V DC	100	5000 RPM
Gray	OV COM	200	2500 RPM
Green	SIG A.	360	1800 RPM
White	SIG B.	500	1200 RPM
Yellow	SIG Z.	1000	500 RPM
Shield	F.G	1024	500 RPM

ORDERING CODE

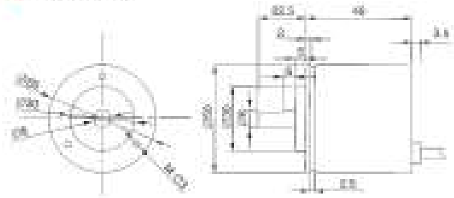
PR-01 - P/R

Pulse / Revolutions	
100	100 P/R
200	200 P/R
360	360 P/R
500	500 P/R
1000	1000 P/R
1024	1024 P/R

TECHNICAL SPECIFICATION

• Supply	: 5-30 VDC, ripple max 3% rms
• Current consumption	: max 40 mA
• Output wave form	: 2-phase + home position (A, B, Z) : Duty ratio 50 ±25% (square wave) : Rise / fall time Max 3 u sec. (max. bi 50 cm)
• Max Speed	: 5000 rpm
• Initial torque	: 0.003 Nm.
• Operating temperature	: 0-50°C
• Operating Vemmidity	: 40-80% RH
• Protection IP 50	: Dust proof
• Vibration resistance	: 10-50 Hz with 0.75 mm amplitude
• Shaft Diameter	: 8 mm.
• Housing Diameter	: 50 mm.
• Cable	: 50 cm, shielded cable

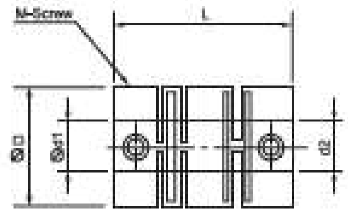
DIMENSION



PR-02

COUPLING

DIMENSION



Model	D1/D2	L	D	Torque	Angular	Material
PR-02-06	06/06	21	15	0.8 N.m.	5 Degree	Polyester Resin
PR-02-08	06/08	25	22	2 N.m.	5 Degree	Polyester Resin

Proximity Switch

Appendix F. Published papers

Design and development of the control system for power ankle prosthesis

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***Abstract*— (*Abstract*) Phase measurement is a key problem in power ankle control system because the traditional control methods is complex mechanical systems and high complicate computation systems. In this article aims to develop of a straightforward phase measurement control system using two Flexi-Force sensor SR1 and SR2, SR1 installed under the toe and sensor SR2 installed under the heel. Experiment with artificial power ankle model in 2 Km/h walking speed. The accelerometer and the encoder will record the experimental result, compared with reference data [12]. The result can shows it can accurately measure the ankle phase during the walking in real-time.**

Keywords— gait analysis, control system, power ankle, Flexi-Force.

I. Introduction

From the first official global report on disability. The proportion of disabled people is rising to 15% of the world's population [1]. How to let the disabled recover the ability to walk is more and more important in future.

Today's commercially available prostheses are no power passive prostheses, it only can get the ability to stance and simple walking in fixed terrain, but the human ankle provides extra power during the walking. The earliest work in powered prostheses was proposed in1972s [2] until today there had so many research about powered prostheses. Samuel K [3-4] use the impedance control of the prosthesis control system, he uses a high power output DC motor, a transmission, a series spring Composition Series-Elastic Actuator (SEA). This SEA can get the torque feedback to the system for control, this prosthesis can mimic the normal human ankle walking. But this prosthesis has to control by PC [5], it cannot work by itself. Frank Sup [6-7] designed a new prosthesis with embedded system [8], it can independently walk in outdoor for 9km.Huseyin Atakan Varol [9] proposed Real-Time Intent Recognition for that prostheses [6, 7 and 10]. This prostheses control system signal is from inertial sensor. Mr.Ugrit Chammar designs a simple mechanical system for the artificial foot [11].It can continuously work about 6000 steps, normal human is 5500 step/day. Therefore it can work all day with full battery.

The disadvantages of that artificial prostheses are impedance control based on high performance computing system and complex mechanical structure. Inertial sensor is difficult to make sure the walking phase in real time, it just estimated movement trend at the beginning of the walking phase, cannot measure every walking phase during the walking. In this paper a straightforward phase estimate control system method based Flexi-Force sensor is developed. It not only can accurately measure the ankle phase during the walking in real-time, but also the signal is easy to calculate.

II. Theory

The level walking gait cycle is defined two parts: stance phase (60% gait cycle) and the swing phase (40% gait cycle) (Fig.1) [3].The stance phase beginning with the heel strike and ending at the top-off , when the foot is off the ground until next heel-strike is swing phase . Different phase has different ankle angle, hell-strike start at 0^0, in stance phase angle varies between -7^0 to 10^0. Therefore phase measurement in real time is important for power ankle control system.

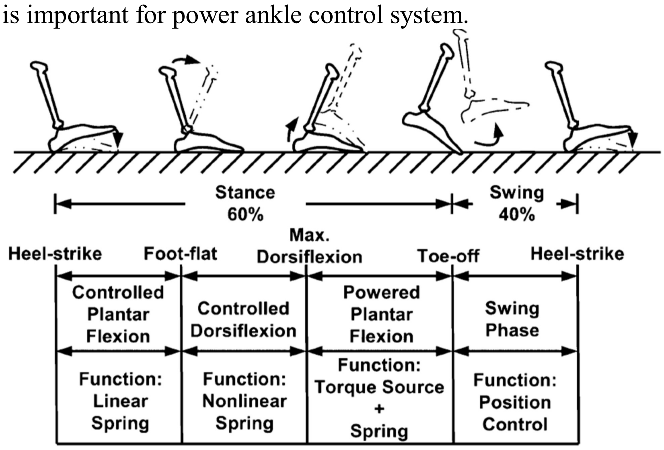

Fig. 1. Gait analysis for level-ground walking.

A. *Walking gait phase measurement using Flexi-force*

In this paper uses two Flexi-force A201Sensor (SR1 and SR2) installed under the sole (Fig.2). Sensor SR1 installed under the fore foot and sensor SR2 installed under the heel. Signals from the SR1 start with a heel-strike when the heel touches the ground and ending at the dorsiflexion when the heel rises from the ground Sensor SR2 start with a foot-flat and ending at the toe off (Fig.3).

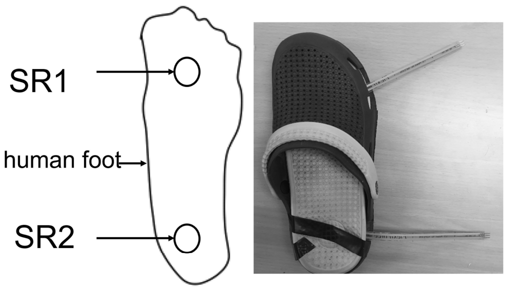

Fig. 2 Flexi-force A201 Sensor position

The sensor shoes was tested on a 27-year-old male (1.86 m, 90 kg) healthy man .The walking data collection from DAQ IN-6009 using LabVIEW Software, sampling rate is 2000/s. Pressure on Sensor is proportional to the voltage. X axis is voltage, Y axis is time. Each step is described as follows.

1) Controlled plantar flexion: begins at heel-strike and ends at the foot-flat. In this phase when heel touch ground sensor SR2 generates the signal.

2) Controlled dorsiflexion: begins at the foot-flat and continues until the ankle maximum angle. In this phase sensor SR1 generates the signal at foot-flat and SR2 stop signal at an ankle maximum angle.

3) Powered plantar flexion: begins at max dorsiflexion and ends at the toe-off phase. In this phase when the toe off the ground the signal SR1 was reduced to zero.

4) Swing phase: begins at after powered plantar flexion until the next heel-strike. In this phase the food was off the ground ,both of the sensor signal SR1 and SR2 was equal to zero.

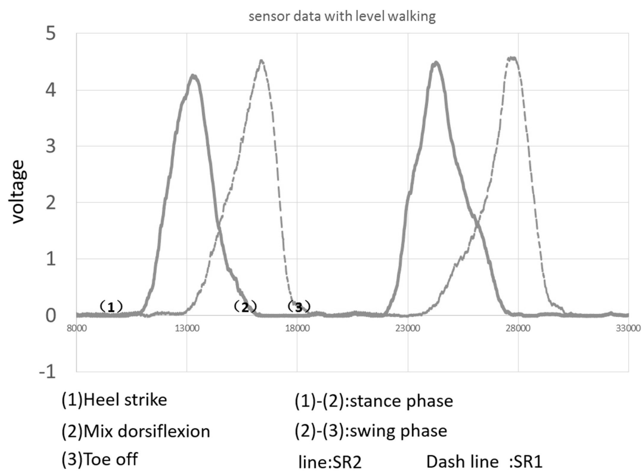

Fig. 3 Flexi-force Sensor data with level walking

B. Walking Controller method with Flexi-force

The walking phase can be Simple and accurate expression by two Flexi-force Sensor SR1 and SR2 (Fig.3). The key question for the power ankle control system is an ankle angle control. A. S. McIntosh measure the ankle angle during the walking on level, 5^0 and 10^0 [12].In this paper, we only use the level-walking data (Fig.9 a).Definition of clockwise rotation will be denoted by a positive and reverse travel as negative, 0^0 is the base when the leg and foot is vertical and threshold is parameter depending on body weight (Fig.4). The controller method is described as follows.

1) Standing to swing phase: system start in the standing and wait sensor signal. When the foot is off the ground, sensor SR1 and SR2 signal equal to zero, in this time reset the ankle angle to zero.

2) Swing phase to max dorsiflexion phase : In this step when the heel touches the ground, SR2 > threshold and SR1 = 0, control the ankle angle from 0^0 to 10^0 and reverse travel to -7^0, delay before the reverse travel for protecting the DC motor drive circuit

3) Powered plantar flexion : In this step the heel is off the ground and the toe still on the ground, SR1 > threshold and SR2 < threshold. Reset the ankle angle to 0^0 and waiting next gait cycle.

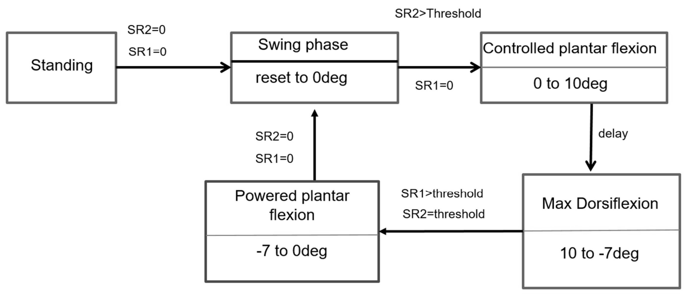

Fig.4 Walking controller with flexi-force signal

III. EXPERIMENTATION AND RESULT

The performance of the controller method with Flexi-force, the experiments using a power ankle model have been performed. Here, we present some illustrative results to show the control capability of the Flexi-force sensor controller system

A. Artificial power ankle model design

In this article, we use plastic to make a simple power ankle model (Fig.5).The left one is a sensor in the shoe to collect the signal (Fig.2), right one is the power ankle model. The model consists of signal processing, power supply, encoder, DC-motor, and DC-motor drives.

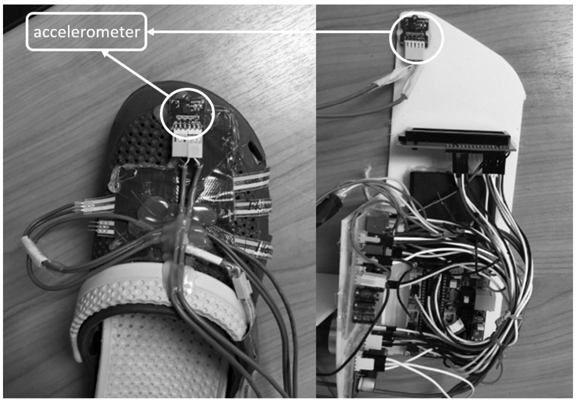

Fig.5 Sensor shoe and power ankle model

This system is powered by 12V and 700mA. The main computational element of the system is 16-bit digital signal controller dspic30f2010.Three-phase encoder can feedback the ankle angle signal to measure the ankle position. Dspic30f2010 consists of 10-bit high-speed analog to digital converter (ADC) modular and quadrature encoder interface (QEI) modular. The analog signal of the sensor can be converted to a digital signal by ADC modular, QEI modular can analysis encoder signal (Fig.6).

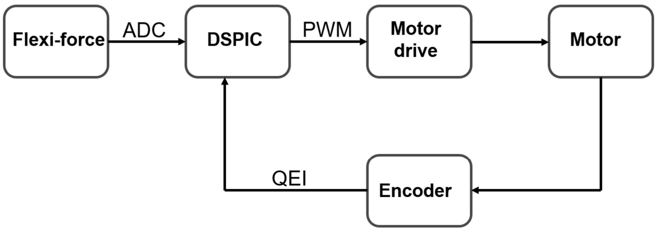

Fig.6 Control system for power ankle model

B. Experimental process

Three axis accelerometer ADXL335 can measurement three directions acceleration XYZ. Z axis vertical to level-ground. Ankle swing will be reflected in Z axis. For validating the validity of control system, accelerometer A1 and A2 respectively installed on sensor shoe and power ankle model (Fig.5).The artificial power ankle is fixed on the thigh, artificial model near with the foot and vertical the leg (Fig.7). Sensor shoes will be worn on the same side of the foot. During the walking, accelerometer A1 and A2 will have two different signals, the two output signals were compared the similarity to control system validation. This experiment is tested on a treadmill and walking speed is 2km/h. Data collection from DAQ IN-6009 using LabVIEW. The voltage on the X axis and Y axis is number of sampling, sampling rate is 2000/s (Fig.8).

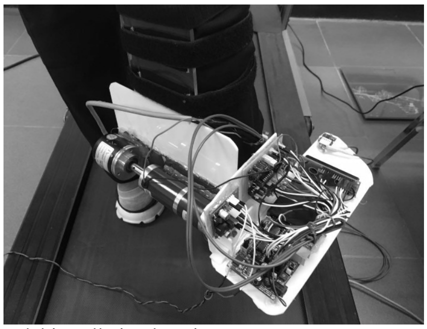
Fig.7 Artificial model installation location

The data collection (Fig.8) of 4 steps walking data, the (Fig.8 a) is a healthy walk signal from A1, (Fig.8 b) is an artificial signal from A2, the walking start in swing phase when heel-strike the heel just touches the ground until the top touches the ground, and there had a great change in Z axis acceleration. During the stance phase Z axis acceleration almost unchanged. During the swing phase when the top is off the ground, the top in the Z axis has a larger swing it will be reflected in the acceleration. From this result healthy and artificial ankle has similar phase change, but it also have noise in the second time swing phase, this noise is from elastic shock in swing phase.

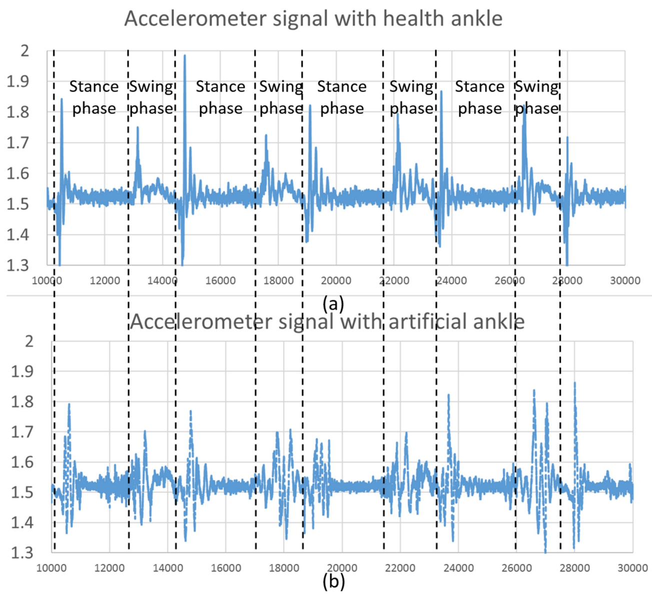

Fig.8 Level-walking result at a speed of 2km/h, a: health ankle b: artificial ankle

In order to better prove that the system is working properly. We also record the ankle angle signal from the encoder and compare with reference data, chart (Fig.9 a) is reference data from [12], and second chart is artificial ankle data (Fig.9 b). Walking phase is described as follows.

1) Heel-strike to foot-flat: 0 to 20 percent, the heel initially make contact with the ground and the top was follow .In this phase ankle angle from 0^0 to 10^0.

2) Foot-flat to max dorsiflexion: 20 to 60 percent , begins at the foot-flat and continues until the ankle maximum angle ankle angle from 10^0 to -7^0.

*3) Swing phase:*60 to 90 percent ,the food off the ground and ankle angle reset to 0^0.

Comparison with reference data [12] from the (Fig.9 a) in (Fig.9).The changing of walking phase is similar, the difference is slope during heel-strike to foot-flat, the problem is from the DC-motor speed more faster than human, this problem can solve it in mechanical system such as gear transmission system .This system can effectively control the ankle angle in each phase during the level-walking.

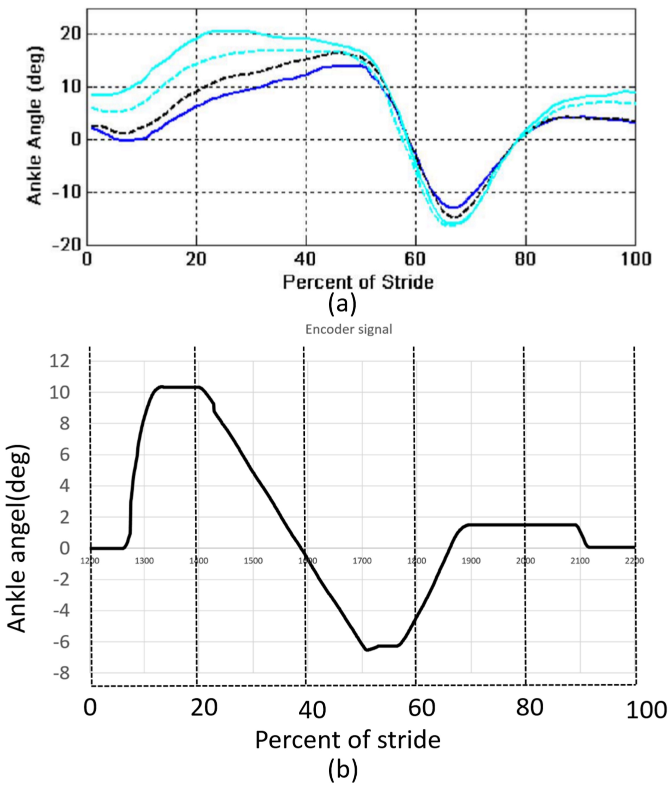

Fig.9 Ankle angle data from level walking.

IV. CONCLUSIONS

In this article, sensor SR1 installed under the fore foot and sensor SR2 installed under the heel, it can accurately measure each walking phase during the walking in real-time. The control system based on the Flexi-force sensor can straightforward measurement the walking phase during the real-time walking. Experiments result showed that is an effective method for power ankle control system. It is more straightforward than traditional control methods impedance control or inertial sensor. The future works are suggest. All of experiment was test on the plastic artificial model as it is east to make. However, the experiment did not give any suggestion that it can work on real artificial power ankle. Therefore, to make a real artificial power ankle and testing this system are necessary. To get this system defect and improve it to perfect the system

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