DYNAMIC RESPONSE PROSTHETIC LEG

DISHA BUDAGE, PRITHVIRAJ GUPTA, HARIPRASAD IYER AND GURINDER SAINI.

ABSTRACT

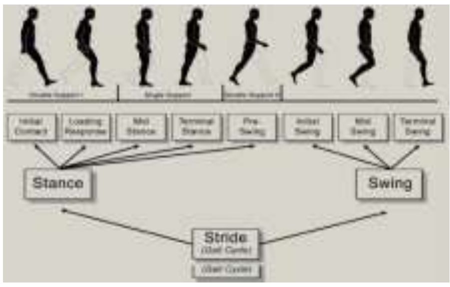
The benefits of a powered prosthesis have been known for a long time. But One of the main hurdles is the challenge of building a reliable below knee ankle prosthesis that is not only similar to the human ankle, but also provides equivalent shock absorbing and gait recognition capabilities. We propose a powered prosthetic ankle concept that overcomes these obstacles . The prosthesis comprises an actuator with motor control and a compression spring. With this architecture, the ankle-foot prosthesis matches the size and weight of the human ankle, and is shown to be satisfying the restrictive design specifications dictated by normal human ankle walking .

INTRODUCTION

A prosthesis or prosthetic implant is an artificial device that replaces a missing body part, which may be lost through trauma, disease, or a condition present at birth (congenital disorder). Prostheses are intended to restore the normal functions of the missing body part. Prostheses can be created by hand or with computer-aided design (CAD), and through various analyses-based software's. During rehabilitation process, the physician assesses the amputee potential level of functional mobility and ability to use lower limb prosthesis and then classifies the lower limb amputees into a range of K-levels .This classification classifies the Trans femoral amputees (TFA) from do not have the ability to ambulate safely without assistance (K0) to who exceeds basic ambulation skills and exhibits performance of high impact stress activities (K4).This type of classification helps prosthetist/physician as guiding rules for prosthetics. Hence it is Important to have a prosthetic with proper functionality that serves its purpose without causing further psychological and physical harm to the patient.

BACKGROUND

GAIT CYCLE



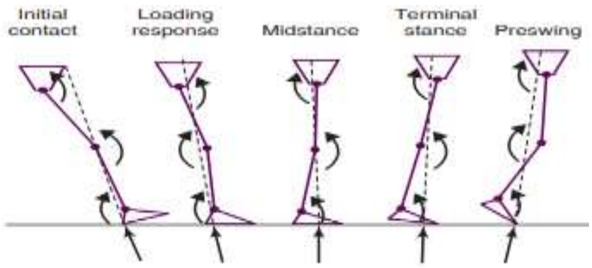
Gait Cycle is a series of movements that form a coherent and energy-efficient motion that results in stable forward propulsion of the body. Gait occurs in different patterns, which are dependent on factors such as the speed of the locomotion required (walking or running). The normal gait cycle consists of two distinct phases (stance and swing), which, for analysis, have been broken down into sub-phases. A single gait-cycle begins at the point at which the foot first touches the ground. When the same foot again makes contact with the ground, a full Gait cycle gets completed. Trauma or disease processes can lead to changes in each of the sub-phases, leading to characteristic and distinct shifts in the pattern of gait.

- GROUND REACTION FORCES DURING THE GAIT CYCLE

As a result of gravity, the weight of the body acts vertically downward on the ground. In accordance with Newton's third law, an equal and opposite force must act upward from the ground on the foot. This is known as the ground reaction force . Obviously, there is no ground reaction force during the swing phase. The ground reaction force has a point of application on the foot, a magnitude, and a direction, or line of action . In static situations these all remain constant, with the magnitude equal to body weight. However, in dynamic situations such as locomotion they typically vary in a repetitive fashion.

The ground reaction force has three components: its point of application, its magnitude, and its line of action. In the stance phase of normal gait, the point of application progresses along the foot, and the magnitude and the line of action vary through the gait cycle. During early and late stance, the magnitude of the ground reaction force is greater than body weight but in midstance is less than body weight as a result of the downward and upward accelerations of the body centre of mass. In normal gait the magnitudes of the two peaks are approximately equal. Importantly, it exceeds body weight in terminal stance.

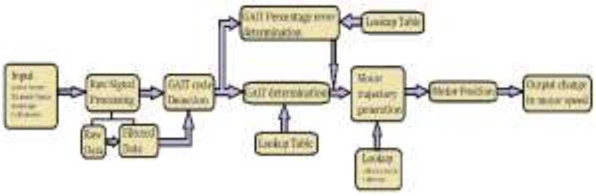
- GAIT PHASE IDENTIFICATION



Due to rise in lower limb amputations ,the use of lower-limb exoskeletons as an alternative for gait rehabilitation, gait phase detection has become an increasingly important feature in the control of these devices. Based on the specific objective of gait phase detection and information contained in the sensor signals, various sensors and algorithms have been developed to classify all or some gait phases. The sensors used in gait phase detection include force sensors, inertial sensors, EMG sensors, Electroneurography (ENG) sensors, and ultrasonic sensors.

PROTOTYPE DEVELPOMENT AND EVALUATION

- CONTROL SYSTEM FOR GAIT ANALYSIS

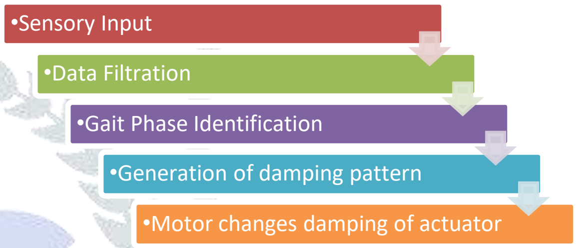


The following control system was made on MATLAB Simulink and evaluated using simulation techniques. Input from various sensors were taken and this raw signal was processed to remove noise from the system. This data was then processed by the gait cycle detection block which identifies the gait phase the patient is currently in. Once the gait phase identification is achieved , subsequent damping is established in the gait determination block. Here the data is compared with pre-set damping values which are derived using mathematical equations. Once the damping values are established ,the change in the motor position is calculated by the motor trajectory generation and is made by the motor position block. Output is subsequent change in the damping.

Such sophisticated systems allow us to decrease the response time of the system , analyse and record the data derived which can further be used to help the patient, validate real world and computer recorded data.

- SOFTWARE

The software/coding behind this project follows the following methodology:



Sensors - ultrasonic and accelerometer are used to collect data. This data is then assigned to specific variables and these variables are then used to calculate gait phase position (stance or swing phases), gait mode (walking, jogging, or running). This information about the gait phase and mode is then used to calculate the suitable damping for the prosthetic ankle.

The program is developed to be used in Arduino and was developed in Arduino IDE.

- SOURCE CODE

```
#include <Stepper.h>
Stepper myStepper(200, 8, 9, 10, 11);
const int pingPin = 7;
const int echoPin = 6;
const int groundpin = 18;
const int powerpin = 19;
const int xpin = A3;
const int ypin = A2;
const int zpin = A1;
void setup()
{
  Serial.begin(9600);
  pinMode(groundpin, OUTPUT);
  pinMode(powerpin, OUTPUT);
  digitalWrite(groundpin, LOW);
  digitalWrite(powerpin, HIGH);
  myStepper.setSpeed(60);
}
long ultrasonic()
{
   long duration, inches, cm;
   pinMode(pingPin, OUTPUT);
   digitalWrite(pingPin, LOW);
   delayMicroseconds(2);
   digitalWrite(pingPin, HIGH);
   delayMicroseconds(10);
   digitalWrite(pingPin, LOW);
   pinMode(echoPin, INPUT);
   duration = pulseIn(echoPin, HIGH);
   inches = duration / 74 / 2;
   cm = duration / 29 / 2;
   return cm;
   delay(100);
}
long accelerometer()
{
  long xspeed,yspeed,zspeed;
  xspeed = analogRead(xpin);
  yspeed = analogRead(ypin);
  zspeed = analogRead(zpin);
  return xspeed;
  delay(100);
}
int differentiate(long velocity)
{
  int mode = 0;
  if (velocity<=2)
    mode = 1;
  else if (velocity<=4)
    mode = 2;
  else if (velocity<=8)
    mode = 3;
  return mode;
}

int gaitphase(long height)
{
  int phase = 0;
  if (height <= 5)
   phase = 1;
  else
   phase = 0;
  return phase;
}
void loop()
{
  long distance;
  distance = ultrasonic();
  Serial.print(distance);
  Serial.print("\n");

  long speed;
  speed = accelerometer();
  Serial.print(speed);
  Serial.print("\n");

 int phase;
  phase = gaitphase(distance);
  if(phase == 0)
   myStepper.step(50);
  else if (phase == 1)
  {
   Serial.print("\nTerminal Swing to Terminal
Stance GAIT Phase");
   int gaitmode;
    gaitmode = differentiate(speed);
    switch (gaitmode)
    {
      case 1:
       myStepper.step(50);
       Serial.print("Walking");
       break;
      case 2:
       myStepper.step(100);
       Serial.print("Jogging");
       break;
      case 3:
       myStepper.step(200);
       Serial.print("Running");
       break;
      default:
       myStepper.step(200);
       break;
   }   }
  delay(250);
}
```

- HARDWARE

List of hardware components used in this project is:

1. Arduino UNO
2. Ultrasonic Distance Sensor - HC-SR04
3. Accelerometer - Sparkfun mma7361 breakout
4. Stepper Motor Nema 17 Hybrid
5. Robodo Drv8825 Stepper Motor Driver
6. MR Damper
7. Connecting Wires and Bread Boards

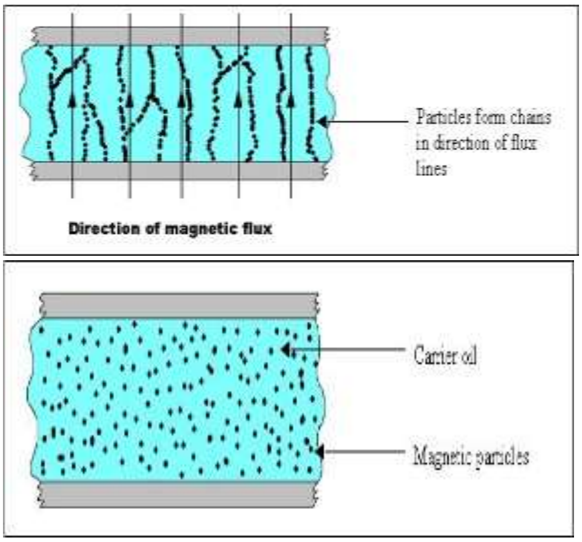
- MOTOR

A stepper motor is a brushless DC electric motor which divides its rotation in several steps. The motor's position can then be commanded to move and hold at one of these steps without any position sensor for feedback, as long as the motor is carefully sized to the application in respect to torque and speed.

The motor used in this project is a 200W BLDC with its dimensions being 10*5.4*5 cm and has a weight of 300 grams with one memory stick capable of changing the motor speed when commanded. Maximum Torque output of the motor is 4 kg-cm with a step angle of 1.8 degrees and 200 steps.

- DAMPING SYSTEM

A **magnetorheological damper** or **magnetorheological shock absorber** is a damper filled with magnetorheological fluid, which is controlled by a magnetic field, usually using an electromagnet. This allows the damping characteristics of the shock absorber to be continuously controlled by varying the power of the electromagnet. Fluid viscosity increases within the damper as electromagnet intensity increases.



<u>Working of a MR damper</u>

The magnetic particles, which are typically micrometre or nanometre scale spheres or ellipsoids, are suspended within the carrier oil and distributed randomly in suspension under normal circumstances. When a magnetic field is applied, however, the microscopic particles (usually in the 0.1–10 µm range) align themselves along the lines of magnetic flux.

1. <u>MR fluid and its properties</u>

To understand the behaviour of the MR fluid it is necessary to model the fluid mathematically, by varying the material properties such as yield stress. Smart fluids are such that they have a low viscosity in the absence of an applied magnetic field, but become quasi-solid with the application of such a field. In the case of MR fluids , the fluid actually assumes properties comparable to a solid when in the activated state, up until a point of yield (the shear stress above which shearing occurs). This yield stress is dependent on the magnetic field applied to the fluid, but will reach a maximum point after which increases in magnetic flux density have no further effect, as the fluid is then magnetically saturated. The behaviour of a MR fluid can thus be considered similar to a Bingham plastic.

Thus, our model of MR fluid behaviour in the shear mode becomes:

$$\mathrm{T} = \tau y(H) + \eta\left(\frac{\eth V}{\partial z}\right), \mathrm{T} > \tau y$$

Where T= shear stress; τy = yield stress; H = Magnetic field intensity; η = Newtonian viscosity; $\frac{\eth V}{\partial z}$ is the velocity gradient in the z direction.

2. <u>Modes of operation</u>

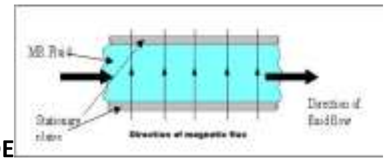
An MR fluid is used in one of three main modes of operation being :

I. Flow mode
II. Shear mode
III. Squeeze-flow mode

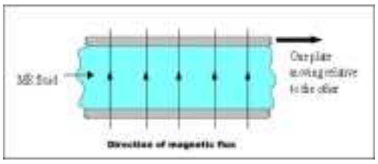
These modes involve fluid flowing as a result of pressure gradient between two stationary plates; fluid between two plates moving relative to one another; and fluid between two plates moving in the direction perpendicular to their planes. In all cases the magnetic field is perpendicular to the planes of the plates, so as to restrict fluid in the direction parallel to the plates.

Hence using MR damper is a good alternative for providing shock absorbing capabilities than a traditional damper system. Using this damper also eliminates the use of a stepper motor which can further decrease the overall response time of the system and increase system efficiency.

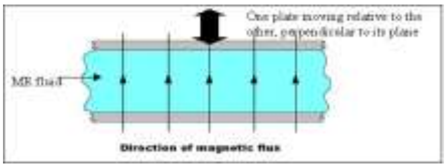
FLOW MODE



SHEAR MODE

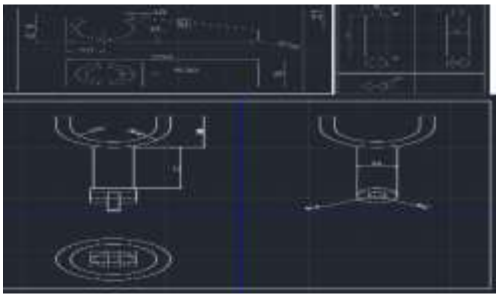
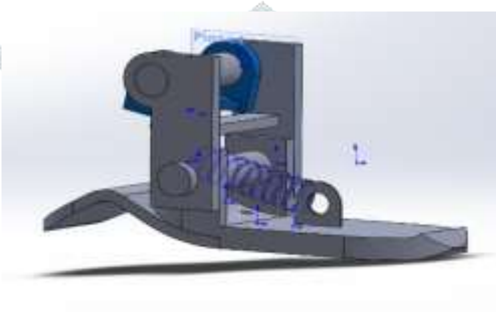
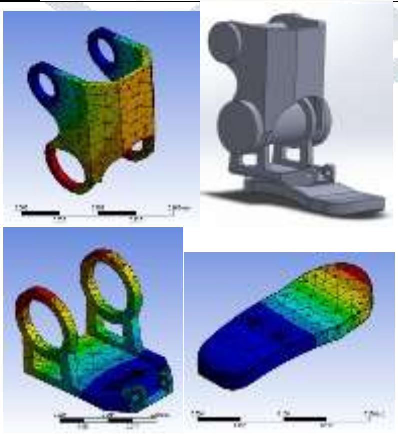


SQUEEZE-FLOW MODE



- MECHANICAL DESIGN

The following table shows the iterative design procedure that was used during the due course of this project. Changes from every previous iteration was made to the next iteration to minimize flaws and maximize efficiency .

Iteration	Image	Remark
Iteration 1		Used for calculating basic design calculations , stresses and initial dimensions.
Iteration 2		3d cad model made with changes from iteration 1 used to evaluate spring force.
Iteration 3		Addition of mechanical damper in the cad model.
Iteration 4		Changes in dimensions from iteration 3 and structural analysis performed on Ansys Mechanical.

- MATERIALS

A variety of metals are used for prosthetics limbs; Aluminium, Titanium, Magnesium, Copper, Steel, and many more. They are each used in a varied amount and for various applications, either pure or alloyed. The use of carbon fibres came about in the 20th century when medics and engineers were in search of a lighter load bearing material. The properties of carbon fibres, such as high stiffness, high tensile strength, low weight, high chemical resistance, high temperature tolerance and low thermal expansion, high specific strength and specific modulus. It was determined that it could be strong enough for even a heavy weight amputee. Materials with high elastic modulus are usually not very ductile: the specific modulus of wood is comparable to that of steel, magnesium, titanium, or aluminium, whereas that of carbon fibre reinforced composites is about three times as high.

Biocompatibility refers to materials that are not harmful to living tissue. An aspect of biocompatibility is how a material interacts with the surface of the skin or the external body. When prosthetics are attached to the exterior of the limb, and constant movement is occurring, the skin can be subject to a variety of painful and uncomfortable side effects. The distribution of mechanical stress at body support interfaces can influence the risk of tissue breakdown. Excessive pressure and shear stress can lead to skin blisters, cysts, or ulceration. Supporting materials used in prosthetics are SpinCo, Poron, Nylon-reinforced silicone, Nickelplast.

CONCLUSION

With this project , a concept of microprocessor powered lower limb ankle prosthesis with variable damping was developed. Microprocessor powered prosthesis has a huge potential in countries with large number of amputees. Ease of use , reparability, functionality, cost, usability ; these factors play a huge role for allowing these types of prosthesis to be used by the masses. In the following project , such a low-cost prosthesis was developed using Computer aided Design and Computer Aided Manufacturing. All parts designed have longer life than its traditional counterparts with more functionality. Addition of a damper and spring system provides better comfort in the long run.

Use of gate mode identification and gate phase identification through sensory input and processing also helps in achieving variable damping of the prosthetic joint which varies according to walking, jogging or running gate as well as according to the phase of gate cycle.

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