DOKUZ EYLÜL UNIVERSITY
GRADUATE SCHOOL OF NATURAL AND APPLIED SCIENCES

MODELLING AND INVESTIGATION OF BIPEDAL HUMAN WALKING SYSTEM

by
Emre SURA

September, 2010
İZMİR

MODELLING AND INVESTIGATION
OF
BIPEDAL HUMAN WALKING SYSTEM

A Thesis Submitted to the
Graduate School of Natural and Applied Sciences of Dokuz Eylül University
In Partial Fulfillment of the Requirements for the Degree of Master of Science
in
Mechanical Engineering, Machine Theory and Dynamics Program

by
Emre SURA

September, 2010
İZMİR

M.Sc THESIS EXAMINATION RESULT FORM

We have read the thesis entitled **“MODELLING AND INVESTIGATION OF BIPEDAL HUMAN WALKING SYSTEM”** completed by **EMRE SURA** under supervision of **PROF. DR. EROL UYAR** and we certify that in our opinion it is fully adequate, in scope and in quality, as a thesis for the degree of Master of Science.

Prof. Dr. Erol UYAR

Supervisor

Yrd. Doç. Dr. Mutlu BOZTEPE

(Jury Member)

Yrd. Doç.Dr. Zeki KIRAL

(Jury Member)

Prof.Dr. Mustafa SABUNCU
Director
Graduate School of Natural and Applied Sciences

ACKNOWLEDGEMENTS

I would like to thank Prof. Dr. Erol UYAR, my thesis supervisor, for his support and guidance. The completion of this dissertation would not be possible without his advice on the research.

Also, I would like to thank my parents, my brothers for their love, support, and patience.

Emre SURA

MODELLING AND INVESTIGATION OF
BIPEDAL HUMAN WALKING SYSTEM

ABSTRACT

In this thesis, bipedal human walking is investigated and its characteristics is explained for modelling and calculating the mechanics of bipedal walking. Firstly phases of walking, then anatomical and mechanical meanings of walking are mentioned to explain principles of bipedal walking. With using Denavit-Hertenberg Homogeneous Transformation Matrices, the mechanics of a human leg and bipedal walking is calculated. This is important for modelling bipedal walking and important to walk amputees as an healthy person again. The types of amputation and solutions of them like prosthesis are mentioned in this thesis. Historical development of the prosthesis is explained also. To understand how these prosthesis works, parts of a prosthetic leg is examined. Classification of prosthetic knees and an example from these prosthetic knee types are explained and described how to control these type of knees. Controlling principles and electronic control program are mentioned in this thesis, and also an electronic control unit is designed.

Keywords: Bipedal human walking, mechanics of walking human leg, amputation, Denavit-Hertenberg Homogeneus Transformation Matrices, prosthesis, prosthetic knee, electronic control of prosthetic knee.

BİPEDAL YÜRÜME DİNAMİĞİNİN ARAŞTIRILMASI VE MODEL TASARIMI

ÖZ

Bu tezde, bipedal yürüme incelenmiş ve bipedal yürümenin modellenmesi ve mekaniğinin hesaplanması için karakteristik özellikleri açıklanmıştır. Bipedal yürümenin prensiplerinin açıklanması için, ilk önce yürümenin fazları , sonra ise yürümenin anatomik ve mekanik anlamları anlatılmıştır. Denavit-Hertenbeg Homojen Transformasyon Matrisleri kullanılarak insan bacağının ve yürümenin mekaniği hesaplanmıştır. Bu, bipedal yürümenin modellenmesinde ve ampute kişlerin tekrar sağlıklı bir insan gibi yürütülmesinde önemlidir. Amputasyonun çeşitleri ve protez gibi çözümleri bu tezde anlatılmıştır. Ayrıca protezlerin tarihsel gelişimi de açıklanmıştır. Bu protezlerin nasıl çalıştıklarının anlaşılabilmesi için bir protez bacağın parçaları incelenmiştir. Prostetik dizlerin sınıflandırılması ve bu prostetik diz çeşitlerinden bir örnek açıklanmış ve bu çeşit dizlerin nasıl kontrol edildiği tanımlanmıştır. Kontrol prensipleri ve elektronik kontrol programı bu tezde anlatılmış, ayrıca bir elektronik kontrol devresi dizaynı yapılmıştır.

Anahtar sözcükler: Bipedal insan yürümesi, yürüyen bir bacağın mekaniği, amputasyon, Denavit-Hertenberg Homojen Transformasyon Matrisleri, protez, prostetik diz, prostetik dizin elektronik kontrolü.

CONTENTS

Page

CHAPTER ONE
INTRODUCTION

1.1 Introduction

There are many researches about designing and manufacturing of walking robots like human in the world. Walking is one of the fundamental movements of human body. Bipedal walking has many advantages if we compare with the mobile wheeled vehicles. If walking is modelled as a pendulum, it needs less input energy to keep it swinging. The hip and ankle joints of an healthy walker move through certain paths. The knee joints of legs make synchronized angular movement during bipedal walking.

1.2 Principle Meaning of Walking

Walking is one of the principal movements of the human body. It is a procedure that is done by consequent steps. The pendulum movement of leg around pelvis, which is made between the times that foot leaves the contacted surface and touches it again is called a step. During this pendulum motion, the other leg contacts the surface and carries all the load of the whole body. When the dynamic leg, which is called the swing leg, passes the static leg (i.e. the stance leg), the body tends to fall forward. But the heels of the swing leg touch the surface so the body automatically preserves its balance. During this movement, reverse swinging arms with the legs help the body to regain its stability.

In the early studies, the 2:1 frequency coordination between arm and leg movements is observed for the normal walking procedure. The occurrence of this frequency coordination has been explained by the tendency of the arms to move as closely as possible to their eigenfrequencies (cf., Craik et al., 1976; Van Emmerik, Wagenaar, & Van Wegen, 1998; Wagenaar & Van Emmerik, 2000; Webb & Tuttle, 1989).

Knee flexion occurs when the leg is bent dorsally (towards the back), whereas extension occurs when the leg is straightened. (Figure 1.1)

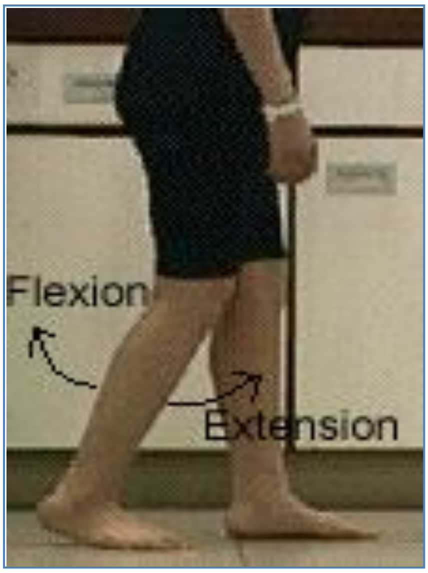

Figure 1.1 Flexion and extension

1.3 Phases of Walking

When we walk, one foot or the other is always in contact with the ground. Each leg is constantly transitioning, going from standing and supporting our weight to swinging through from behind to in front of us to get ready for the next step. The legs are always transitioning from stance to swing, which is why our walking motion is divided into what we call the "swing phase" and the "stance phase." (Figure 1.2)

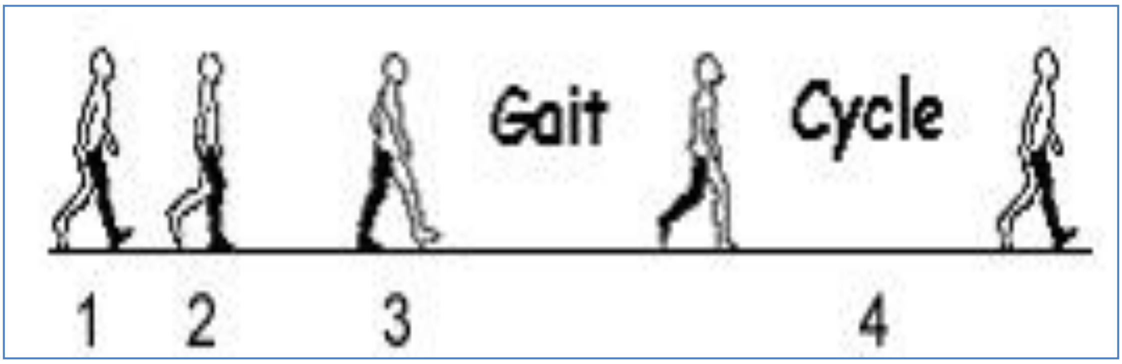

Figure 1.2 Gait cycle

1.3.1 Stance Phase

The stance phase is from the foot contacts with the ground until it rises from the ground. It takes approximately 60 percent of the gait cycle.

The stance phase divided in three phases. (Figure 1.3)

1.3.1.1 Initial Contact and Heel Strike

The stance phase begins with heel strike shown as 1 in figure. This phase commences when the heel strikes the ground with the leg in full extension, and progresses through a few degrees of flexion. The period ends when the forefoot makes contact with ground and it lasts for about 25 percent of stance phase.

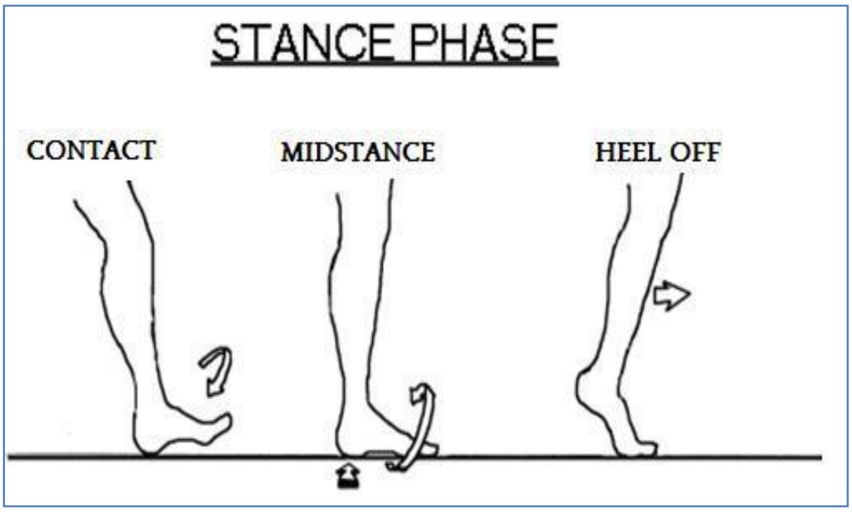

Figure 1.3 The stance phase of walking

1.3.1.2 Mid Stance

In mid stance, the leg and foot are in stable and the center of gravity is directly over the foot and the knee is in full extension. It is shown as 2 in figure. In this phase, the other leg is in swing phase so that all weight of human body effects on stance foot. This period lasts for about 50 percent of stance phase.

1.3.1.3 Terminal Stance and Heel Off

This phase is final stage of stance phase and continues until the center of gravity is directly over the contralateral foot and initial foot lifts off the ground. When just the tips of your toes are touching the ground behind you, you've reached the end of the stance phase.

After the stance phase ends, now you're transitioning into the swing phase.

1.3.2 Swing Phase

The swing phase is the time when the foot is in air. It takes approximately 40 percent of the gait cycle. The swing phase begins when the foot is lifted from the floor until the heel is placed down. While walking the thorax rotates in clockwise and counterclockwise directions opposite the pelvic rotations. Some people display more rotation of the thorax, while others display more rotation of the pelvis. With each step the pelvis drops a few degrees on the side of the non-weight bearing, or swinging, leg. While the leg is swinging, the hip abductors of the weight bearing leg contract in order to prevent the pelvis from falling excessively on the unsupported side.

The swing phase divided in four phases.

1.3.2.1 Initial Swing

This part of swing phase is from the toe heel off to opposite foot in stance phase. It begins the moment the foot leaves the ground and continues until maximum knee flexion occurs, when the swinging extremity is directly under the body and directly opposite the stance limb. (Radu & Baritz, 2007)

1.3.2.2 Mid Swing

The mid swing phase is from end of initial swing to the swing limb is in front of the body and the tibia is vertical.

1.3.2.3 Terminal Swing

The terminal swing phase begins tibia is vertical and ends until the foot contacts with ground.

1.4 Anatomical Meaning of Walking

Walking is one of an everyday task that we do and we do not think about how it occurs anatomically. In truth, these tasks require a sophisticated sequence of activities that is impressive. The nervous system provides the pathways to permit us to carry out such precise activities. To understand how it is able to exert such perfect control on our bodies, we have to examine neurons, the most basic part of the nervous system and to consider the way in which nerve impulses are transmitted throughout the brain and the body.

If we consider walking as a swift working mechanism, brain is the control system (regulator in a control system) of human body and neurons are the electrical wires through which the brain gets and sends messages. These messages are purely electrical and neurons follow an all-or-none law, which means they are either on or off; once triggered beyond a certain point, they will be fired. Messages from the brain reach muscles with the help of these electrical wires.

1.5 Mechanical Meaning of Walking

In terms of engineering, walking is a multi degrees of freedom mechanism that has one joint at hip, one joint at knee and one joint in ankle working with an equivalent counterpart with a phase. The friction force, which is caused by the interaction of the soles with the walking surface, provides the movement of the body like the tires of an automobile. Bones are rigid elements that carry the bodyweight and muscles are elastic actuators that drive the bones.

With a traditional mechanism with the known values of leg and body weights, inertias and specific values of speed of the mass centre of the body, it is theoretically possible to consider walking as a mechanism and analyse it. However, it must be remembered that the system is a non-linear system in nature and so in practical applications, it is very difficult to have exact results. Hence, we have to make advanced dynamics approaches to understand the meaning of walking.

It is commonly thought that walking is an active process, that is, some complex pattern of muscular activity is required to produce the motion. Actually, however, walking can be sustained passively by a simple interaction of gravity and inertia. We have built a machine which demonstrates the effect. It is actually little more than two rigid legs connected by a pin joint. If left standing upright with legs together, the machine topples like a pencil on its point. (It can only topple longitudinally; sideways motion is prevented by building each leg as a pair of crutches connected by a rigid link.) However if placed on a shallow downhill slope (which provides a source of energy) and given appropriate initial conditions, it settles after a few steps into a steady gait quite comparable to human walking. Passive dynamic effects both generate and stabilise the gait; the only active intervention is for lifting of the swing feet to prevent toe-stubbing.

Passive walking provides a simple but rigorous model of human locomotion, as well as a foundation for design of practical bipedal machines.

CHAPTER TWO
MECHANICS OF
BIPEDAL HUMAN WALKING

2.1 Introduction

We can think movement of a human leg similar as a planar 3-dof manipulator's movement. We can calculate the position of foot and also we can calculate positions of members of a human leg and the forces which is needed to move a human leg with Denavit-Hartenberg Homogeneous Transformation Matrices. In this chapter we will explain how can we use these matrices to calculate these positions and forces.

2.2 Denavit-Hertenberg Homogeneous Transformation Matrices

Having established a coordinate system to each link of a manipulator, a 4x4 transformation matrix relating two successive coordinate systems can be established. Observation of Figure 2.1 reveals that the i th coordinate system can be thought of as being displaced from the (i-1)th coordinate system by the following successive rotations and translations.

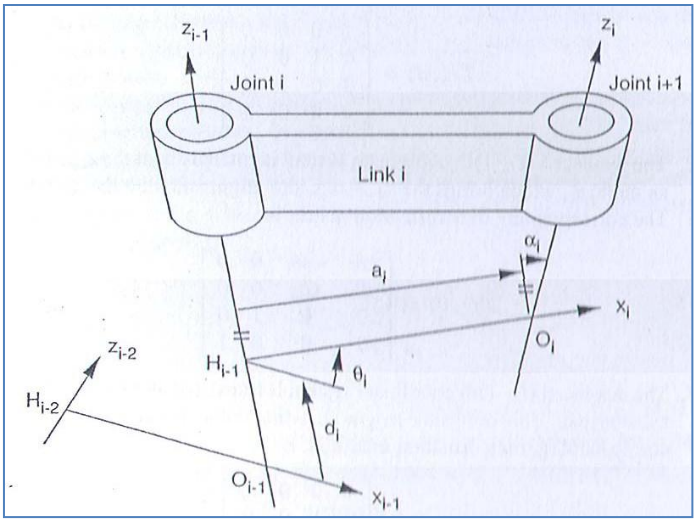

Figure 2.1 Definition of link parameters

1. The (i-1)th coordinate system is translated along the z_{i-1}-axis a distance d_i. This brings the origin O_{i-1} into coincidence with H_{i-1}. The corresponding transformation matrix is

$$T(z,d) = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & d_i \\ 0 & 0 & 0 & 1 \end{bmatrix}$$

2. The displaced (i-1)th coordinate system is rotated about the z_{i-1}-axis an angle θ_i , which brings the x_{i-1}-axis into alignment with x_i-axis. The corresponding transformation matrix is

$$T(z,\theta) = \begin{bmatrix} \cos\theta_i & -\sin\theta_i & 0 & 0 \\ \sin\theta_i & \cos\theta_i & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}$$

3. The displaced (i-1)th coordinate system is translated along the x_i-axis a distance a_i. This brings the origin O_{i-1} into coincidence with O_i. The corresponding transformation matrix is

$$T(x,a) = \begin{bmatrix} 1 & 0 & 0 & a_i \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}$$

4. The displaced (i-1)th coordinate system is rotated about the x_i-axis an angle α_i, which brings the two coordinate systems into complete coincident. The corresponding transformation matrix is

$$T(x,\alpha) = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & \cos\theta_i & -\sin\theta_i & 0 \\ 0 & \sin\theta_i & \cos\theta_i & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix}$$

We may think of the transformations above as four basic transformations about the moving coordinate axes. Therefore, the resulting transformation matrix, ${}^{i-1}A_i$, is given by

$$ {}^{i-1}A_i = T(z,d) \cdot T(z,\theta) \cdot T(x,a) \cdot T(x,\alpha) \tag{2.1} $$

Expanding Eq. (2.1), we obtain

$$ {}^{i-1}A_i = \begin{bmatrix} \cos\theta_i & -\cos\alpha_i \sin\theta_i & \sin\alpha_i \sin\theta_i & a_i \cos\theta_i \\ \sin\theta_i & \cos\alpha_i \cos\theta_i & -\sin\alpha_i \cos\theta_i & a_i \sin\theta_i \\ 0 & \sin\alpha_i & \cos\alpha_i & d_i \\ 0 & 0 & 0 & 1 \end{bmatrix} \tag{2.2} $$

Equation (2.2) is called the Denavit-Hartenberg (D-H) transformation matrix. The trailing subscript i and the leading subscript i-1 denote that the transformation takes place from the i^{th} coordinate system to the (i-1)th coordinate system.

Let the homogenous coordinates of the position vector of a point relative to the i^{th} coordinate system be denoted by ${}^{i}p = [\, p_x , p_y , p_z , 1]^T$. Also let the homogenous coordinates of a unit vector expressed in the i^{th} coordinate system be denoted by ${}^{i}u = [\, u_x , u_y , u_z , 0]^T$. Then the transformation of a position vector and a unit vector from the i^{th} to the (i-1)th coordinate system can be written as

$$ {}^{i-1}p = {}^{i-1}A_i \cdot {}^{i}p \tag{2.3} $$

$$ {}^{i-1}u = {}^{i-1}A_i \cdot {}^{i}u \tag{2.4} $$

Note that the leading superscript is used to indicate system with respect to which a vector is expressed. Although the transformation matrix A is not orthogonal, the inverse transformation exists and is given by

$$^{i-1}A_i = (^{i-1}A_i)^{-1} = \begin{bmatrix} \cos\theta_i & \sin\theta_i & 0 & -a_i \\ -\cos\alpha_i \sin\theta_i & \cos\alpha_i \cos\theta_i & \sin\alpha_i & -d_i \sin\theta_i \\ \sin\alpha_i \sin\theta_i & -\sin\alpha_i \cos\theta_i & \cos\alpha_i & -d_i \cos\alpha_i \\ 0 & 0 & 0 & 1 \end{bmatrix} \qquad (2.5)$$

If we can say a human leg as a planar 3-DOF manipulator, we can explain and calculate the positions of its members and its forces with an example like this.

2.3 Example for Planar 3-DOF Manipulator

Figure 2.2 shows a 3-dof planar manipulator constructed with three revolute joints located at points O_0, A and P, respectively. A coordinate system is attached to each linki The (x_0, y_0, z_0) coordinate system is attached to the base with its origin located at the first joint pivot and x-axis pointing to the right. Since the joint axes are parallel to each other, all the twist angles α_i and translational distances d_i are zero.

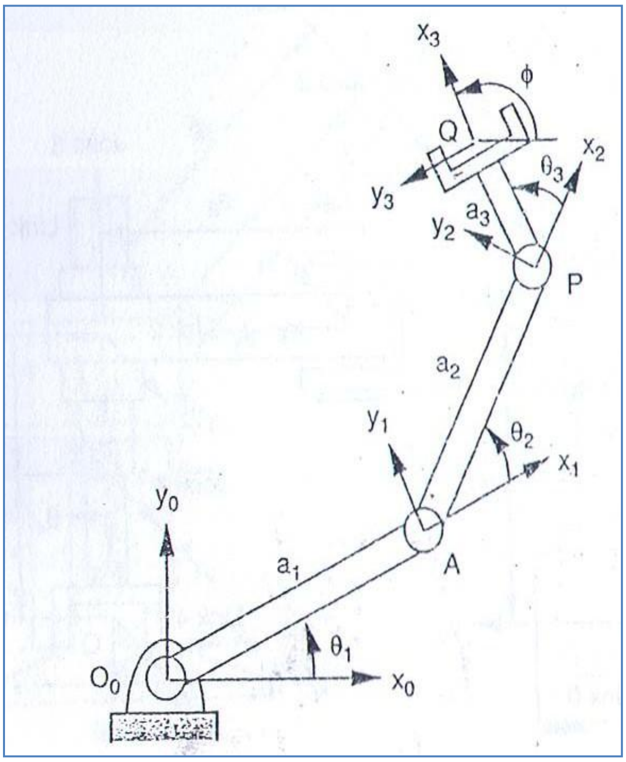

Figure 2.2 Planar 3-dof manipulator

For the coordinate systems chosen the link parameters are given in Table 2.1. The D-H transformation matrices are obtained by substituting the D-H link parameters into Eq.(2.2):

$$^{0}A_1 = \begin{bmatrix} \cos\theta_1 & -\sin\theta_1 & 0 & a_1\cos\theta_1 \\ \sin\theta_1 & \cos\theta_1 & 0 & a_1\sin\theta_1 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2.6)$$

$$^{1}A_2 = \begin{bmatrix} \cos\theta_2 & -\sin\theta_2 & 0 & a_2\cos\theta_2 \\ \sin\theta_2 & \cos\theta_2 & 0 & a_2\sin\theta_2 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2.7)$$

$$^{2}A_3 = \begin{bmatrix} \cos\theta_3 & -\sin\theta_3 & 0 & a_3\cos\theta_3 \\ \sin\theta_3 & \cos\theta_3 & 0 & a_3\sin\theta_3 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2.8)$$

Table 2.1 D-H Parameters of a 3-DOF Manipulator

Joint i	**α_i**	**a_i**	**d_i**	**θ_i**
1	0	a_1	0	θ_1
2	0	a_2	0	θ_2
3	0	a_3	0	θ_3

2.3.1 Position Analysis of a Planar 3-DOF Manipulator

For the planar 3-dof manipulator shown in Figure 2.2, the overall transformation matrix is given by

$$^{0}A_3 = {}^{0}A_1 \cdot {}^{1}A_2 \cdot {}^{2}A_3 \quad (2.9)$$

Substituting Eq. (2.6) through (2.8) into (2.9), we obtain

$$^{0}A_3=\begin{bmatrix}\cos\theta_{123} & -\sin\theta_{123} & 0 & a_1\cos\theta_1+a_2\cos\theta_{12}+a_3\cos\theta_{123}\\ \sin\theta_{123} & \cos\theta_{123} & 0 & a_1\sin\theta_1+a_2\sin\theta_{12}+a_3\sin\theta_{123}\\ 0 & 0 & 1 & 0\\ 0 & 0 & 0 & 1\end{bmatrix} \qquad (2.10)$$

2.3.1.1 Direct Kinematics

The position vector of the origin Q expressed in the end-effector coordinate system is given by

$$^{3}q = [0, 0, 0, 1]^T$$

Let the position vector of Q with respect to the base coordinate system be

$$^{0}q = [q_x, q_y, q_z, 1]^T$$

Then we can relate ^{3}q to ^{0}q by the following transformation:

$$\begin{bmatrix}q_x\\ q_y\\ q_z\\ 1\end{bmatrix} = {}^{0}A_3 \cdot \begin{bmatrix}0\\0\\0\\1\end{bmatrix} = \begin{bmatrix}a_1\cos\theta_1+a_2\cos\theta_{12}+a_3\cos\theta_{123}\\ a_1\sin\theta_1+a_2\sin\theta_{12}+a_3\sin\theta_{123}\\ 0\\ 1\end{bmatrix} \qquad (2.11)$$

Hence, given θ_1, θ_2, and θ_3, the position of point Q can be computed by Eq. (2.11). Similarity, the position vector of any other point in the end effector,

$$^{3}g = [g_u, g_v, 0, 1]^T$$

$$\begin{bmatrix} g_x \\ g_y \\ g_z \\ 1 \end{bmatrix} = {}^0A_3 \begin{bmatrix} g_u \\ g_v \\ 0 \\ 1 \end{bmatrix} = \begin{bmatrix} g_u\cos\theta_{123} - g_v\sin\theta_{123} + a_1\cos\theta_1 + a_2\cos\theta_{12} + a_3\cos\theta_{123} \\ g_u\sin\theta_{123} - g_v\cos\theta_{123} + a_1\sin\theta_1 + a_2\sin\theta_{12} + a_3\sin\theta_{123} \\ 0 \\ 1 \end{bmatrix} \quad (2.12)$$

From Eq. (2.10), we conclude that the orientation angle of the end effector is equal to $\theta_1+\theta_2+\theta_3$.

2.3.1.2 Inverse Kinematics

For the inverse kinematics problem, the location of the end-effector is given and the problem is to find the joint angles θ_i , i = 1, 2, 3, necessary to bring the end-effector to the desired location. For a planar 3-dof manipulator, the end-effector can be specified in terms of the position of point Q and an orientation angle ϕ of the end-effector. Hence the overall transformation matrix from the end-effector coordinate system to the base coordinate system, ${}^0A_3$, is given by

$${}^0A_3 = \begin{bmatrix} \cos\phi & -\sin\phi & 0 & q_x \\ \sin\phi & \cos\phi & 0 & q_y \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2.13)$$

Inverse kinematics solutions can be obtained by equating the elements of Eq. (2.10) to that of (2.13). To find the orientation of the end-effector, we equate the (1,1) and (2,1) elements of Eq.(2.10) to that of (2.13):

$$\cos\theta_{123} = \cos\phi \quad (2.14)$$

$$\sin\theta_{123} = \sin\phi \quad (2.15)$$

Hence;

$$\theta_{123} = \theta_1 + \theta_2 + \theta_3 = \phi \quad (2.16)$$

Next, we equate the (1,4) and (2,4) elements of Eq. (2.10) to that of (2.13):

$$p_x = a_1 \cos\theta_1 + a_2 \cos\theta_{12} \tag{2.17}$$

$$p_y = a_1 \sin\theta_1 + a_2 \sin\theta_{12} \tag{2.18}$$

where $p_x = q_x - a_3 \cos\phi$ and $p_y = q_y - a_3 \sin\phi$ denote the position vector of the point P located at the third joint axis shown in Figure 2.2. Note that by using this substitution, θ_3 disappears from Eq. (2.17) and (2.18). From Figure 2.2 we observe that the distance from point O to P is independent of θ_1. Hence we can eliminate θ_1 by summing the squares of Eq. (2.17) and (2.18); that is,

$$p_x^2 + p_y^2 = a_1^2 + a_2^2 + 2\, a_1\, a_2 \cos\theta_2 \tag{2.19}$$

Solving Eq. (2.19) for θ_2, we obtain

$$\Theta_2 = \cos^{-1}\kappa, \tag{2.20}$$

where

$$\kappa = \frac{p_x^2 + p_y^2 - a_1^2 - a_2^2}{2a_1a_2}$$

Equation (2.20) yields (1) two real roots if $|\kappa| < 1$, (2) one double root if $|\kappa| = 1$, and (3) no real roots if $|\kappa| > 1$. In general , if $\theta_2 = \theta_2^*$ is a solution, $\theta_2 = -\theta_2^*$ is also a solution, where $\pi \geq \theta_2^* \geq 0$. We call $\theta_2 = \theta_2^*$ the elbow-down solution and $\theta_2 = -\theta_2^*$ the elbow-up solution. If $|\kappa| = 1$, the leg is in a fully stretched or folded configuration. If $|\kappa| > 1$, the position is not reachable.

Corresponding to each θ_2, we can solve θ_1 by expanding Eq.(2.17) and (2.18) as follows:

$$(a_1 + a_2\cos\theta_2)\cos\theta_1 - (a_2\sin\theta_2)\sin\theta_1 = p_x \qquad (2.21)$$

$$(a_2\sin\theta_2)\cos\theta_1 + (a_1 + a_2\cos\theta_2)\sin\theta_1 = p_y \qquad (2.22)$$

Solving Eq. (2.21) and (2.22) for $\cos\theta_1$ and $\sin\theta_1$, yields

$$\cos\theta_1 = \frac{p_x (a_1 + a_2 \cos\theta_2) + p_y}{\Delta},$$

$$\sin\theta_1 = \frac{-p_x a_2\sin\theta_2 + p_y(a_1+a_2\cos\theta_2)}{\Delta}$$

where $\Delta = a_1^2 + a_2^2 + 2a_1a_2\cos\theta_2$. Hence, corresponding to each θ_2, we obtain a unique solution for θ_1:

$$\theta_1 = \arctan2(\sin\theta_1,\cos\theta_1) \qquad (2.23)$$

In a computer program we may use the function arctan2(x,y) to obtain a unique solution for θ_1. However, the solution may be real or complex. A complex solution corresponds to an end-effector locatio that is not reachable by the manipulator. Once θ_1 and θ_2 are known, Eq. (2.16) yields a unique solution for θ_3. Hence, corresponding to a given end-effector location, there are generally two real inverse kinematics solutions, one being the reflection of the other about a line connecting points O and P, as illustrated Figure 2.3.

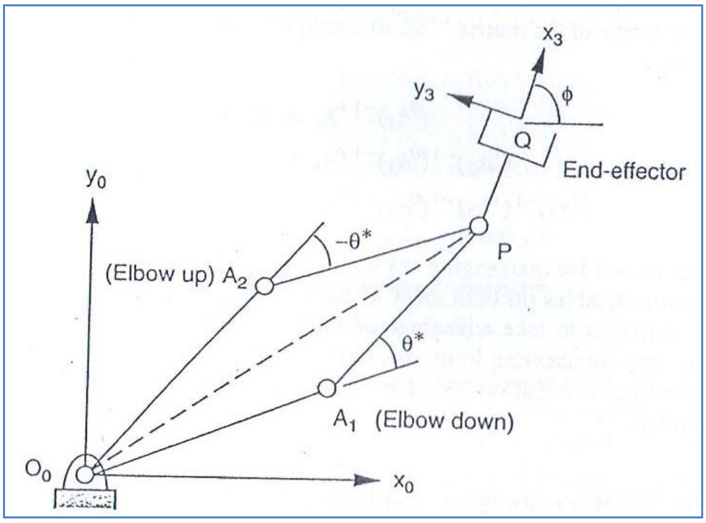

Figure 2.3 Two possible inverse kinematics solutions

2.3.2 Jacobian of a Planar 3-DOF Manipulator

We first compute the vectors z_{i-1} and ${}^{i-1}p_3{}^*$ from Eq.(2.24) , (2.25) , (2.26) and (2.27), for i = 1,2 and3 as follows:

$$z_0 = z_1 = z_2 = \begin{bmatrix} 0 \\ 0 \\ 1 \end{bmatrix} \tag{2.24}$$

$${}^{2}p_3{}^* = \begin{bmatrix} a_3\cos\theta_{123} \\ a_3\sin\theta_{123} \\ 0 \end{bmatrix} \tag{2.25}$$

$${}^{1}p_3{}^* = \begin{bmatrix} a_2\cos\theta_{12} + a_3\cos\theta_{123} \\ a_2\sin\theta_{12} + a_3\sin\theta_{123} \\ 0 \end{bmatrix} \tag{2.26}$$

$${}^{1}p_3{}^* = \begin{bmatrix} a_1\cos\theta_1 + a_2\cos\theta_{12} + a_3\cos\theta_{123} \\ a_1\sin\theta_1 + a_2\sin\theta_{12} + a_3\sin\theta_{123} \\ 0 \end{bmatrix} \tag{2.27}$$

where $\theta_{12} = \theta_1 + \theta_2$ and $\theta_{123} = \theta_1 + \theta_2 + \theta_3$. Substiuting the expressions above into this equation,

$$\begin{bmatrix} v_x \\ v_y \\ w_z \end{bmatrix} = J \begin{bmatrix} \dot{\theta}_1 \\ \dot{\theta}_2 \\ \dot{\theta}_3 \end{bmatrix}$$

where

$$J=\begin{bmatrix} -(a_1 \sin\theta_1 + a_2 \sin\theta_{12} + a_3 \sin\theta_{123}) & -(a_2 \sin\theta_{12} + a_3 \sin\theta_{123}) & -a_3 \sin\theta_{123} \\ (a_1 \cos\theta_1 + a_2 \cos\theta_{12} + a_3 \cos\theta_{123}) & (a_2 \cos\theta_{12} + a_3 \cos\theta_{123}) & a_3 \cos\theta_{123} \\ 1 & 1 & 1 \end{bmatrix} \tag{2.28}$$

We note that if the reference point is chosen at origin of the (x_2,y_2) frame, the Jacobian matrix reduces to

$$J = \begin{bmatrix} -(a_1 \sin\theta_1 + a_2 \sin\theta_{12}) & -(a_2 \sin\theta_{12}) & 0 \\ (a_1 \cos\theta_1 + a_2 \cos\theta_{12}) & (a_2 \cos\theta_{12}) & 0 \\ 1 & 1 & 1 \end{bmatrix} \tag{2.29}$$

2.3.3 Statics of a Planar 3-DOF Manipulator

A coordinate system with all the z-axis pointing out of the paper is defined for each link according to the D-H convention. Let the end-effector output force and moment be given by $f_{4,3} = [f_x, f_y, 0]^T$ and $n_{4,3} = [0, 0, n_z]^T$, respectively.Also let the acceleration of gravity, g, be pointing along the negative y_0-direction and the center of mass be located at the midpoint of each link. We wish to find the joint reaction forces and moments.

The D-H parameters and transformation matrices are given in Table 2.1 and Equation 2.6 through 2.8 . The vectors ${}^{i}r_i$ and ${}^{i}r_{ci}$ are

$$ {}^{i}r_i = \begin{bmatrix} a_i \\ 0 \\ 0 \end{bmatrix} \text{ and } {}^{i}r_{ci} = \begin{bmatrix} -a_i/2 \\ 0 \\ 0 \end{bmatrix} \qquad (2.30) $$

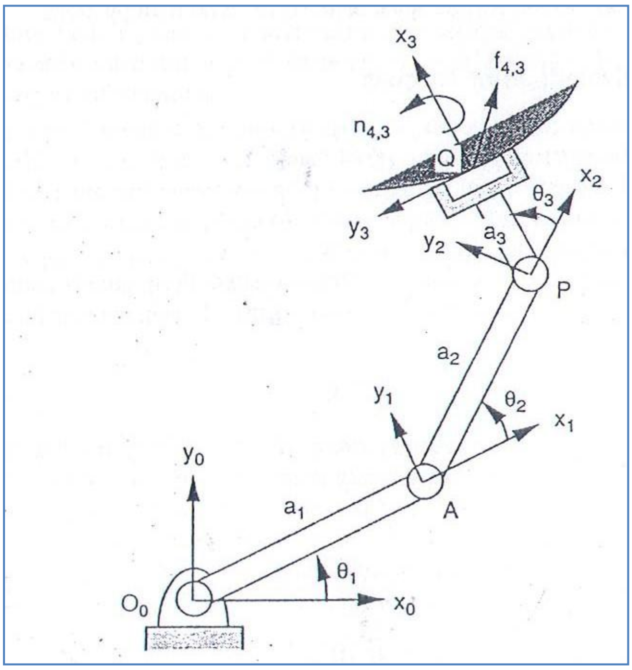

Figure 2.4 Planar 3R manipulator exerting a force $f_{4,3}$ and a moment $n_{4,3}$.

$$ r_1 = {}^{0}R_1\, {}^{1}r_1 = a_1 \begin{bmatrix} \cos\theta_1 \\ \sin\theta_1 \\ 0 \end{bmatrix} , \quad r_{c1} = {}^{0}R_1\, {}^{1}r_{c1} = -\frac{a_1}{2} \begin{bmatrix} \cos\theta_1 \\ \sin\theta_1 \\ 0 \end{bmatrix}, $$

$$ r_2 = {}^{0}R_2\, {}^{2}r_2 = a_2 \begin{bmatrix} \cos\theta_{12} \\ \sin\theta_{12} \\ 0 \end{bmatrix} , \quad r_{c2} = {}^{0}R_2\, {}^{2}r_{c2} = -\frac{a_2}{2} \begin{bmatrix} \cos\theta_{12} \\ \sin\theta_{12} \\ 0 \end{bmatrix}, $$

$$ r_3 = {}^{0}R_3\, {}^{3}r_3 = a_3 \begin{bmatrix} \cos\theta_{123} \\ \sin\theta_{123} \\ 0 \end{bmatrix}, \quad r_{c3} = {}^{0}R_3\, {}^{1}r_{c3} = -\frac{a_3}{2} \begin{bmatrix} \cos\theta_{123} \\ \sin\theta_{123} \\ 0 \end{bmatrix} $$

We now apply Equations;

$$f_{i,i-1} = f_{i+1,i} - m_i g \qquad (2.31)$$

$$n_{i,i-1} = n_{i+1,i} + r_i \text{ x } f_{i,i-1} - r_{ci} \text{ x } m_i g \qquad (2.32)$$

to compute the reaction forces exerted on link 3, then proceed to link 2 and 1 in sequence. For i = 3, substituting r_3 , r_{c3} , $f_{4,3}$ and $n_{4,3}$ into Equations (2.31) and (2.32) yields

$$f_{3,2} = f_{4,3} - m_3 g = \begin{bmatrix} f_x \\ f_y + m_3 g_c \\ 0 \end{bmatrix},$$

$$n_{3,2} = n_{4,3} + r_3 \text{ x } f_{3,2} - r_{c3} \text{ x } m_3 g = \begin{bmatrix} 0 \\ 0 \\ n_{3,2z} \end{bmatrix},$$

where

$$n_{3,2z} = n_z + f_y a_3 \cos\theta_{123} - f_x a_3 \sin\theta_{123} + 0.5\, m_3 g_c a_3 \cos\theta_{123}$$

For i =2, we subsititute $f_{3,2}$ and $n_{3,2}$ obtained in the preceding step along with r_2 and r_{c2} into Equations (2.31) and (2.32) . As a result, we obtain

$$f_{2,1} = f_{3,2} - m_2 g = \begin{bmatrix} f_x \\ f_y + (m_2 + m_3) g_c \\ 0 \end{bmatrix},$$

$$n_{2,1} = n_{3,2} + r_2 \text{ x } f_{2,1} - r_{c2} \text{ x } m_2 g = \begin{bmatrix} 0 \\ 0 \\ n_{2,1z} \end{bmatrix},$$

where

$$n_{2,1z} = n_z + f_y(a_2 \cos\theta_{12} + a_3 \cos\theta_{123}) - f_x(a_2 \sin\theta_{12} + a_3 \sin\theta_{123})$$
$$+ 0.5\, m_2\, g_c\, a_2 \cos\theta_{12} + m_3\, g_c\, (a_2 \cos\theta_{12} + 0.5\, a_3 \cos\theta_{123})$$

For i=1, we substitute $f_{2,1}$ and $n_{2,1}$ obtained in the preceding step along with r_1 and r_{c1} into Equations (2.31) and (2.32). This produces

$$f_{1,0} = f_{2,1} - m_1 g = \begin{bmatrix} f_x \\ f_y + (m_1+m_2+m_3)g_c \\ 0 \end{bmatrix},$$

$$n_{1,0} = n_{2,1} + r_1 \times f_{1,0} - r_{c1} \times m_1 g = \begin{bmatrix} 0 \\ 0 \\ n_{1,0z} \end{bmatrix},$$

where

$$n_{1,0z} = n_z + f_y(a_1 \cos\theta_1 + a_2 \cos\theta_{12} + a_3 \cos\theta_{123}) - f_x(a_1 \sin\theta_1 + a_2 \sin\theta_{12} + a_3 \sin\theta_{123})$$
$$+ 0.5\, m_1\, g_c\, a_1 \cos\theta_1 + m_2\, g_c\, (a_1 \cos\theta_1 + 0.5 a_2 \cos\theta_{12})$$
$$+ m_3\, g_c\, (a_1 \cos\theta_1 + a_2 \cos\theta_{12} + 0.5\, a_3 \cos\theta_{123})$$

Finally, we apply Equation

$$\tau_i = z_{i-1}^T\, n_{i,i-1} \qquad (2.33)$$

to compute the joint torques as follows:

$$\tau_1 = z_0^T\, n_{1,0} = n_{1,0z},$$

$$\tau_2 = z_1^T\, n_{2,1} = n_{2,1z},$$

$$\tau_3 = z_2^T\, n_{3,2} = n_{3,2z}.$$

We note that in the absence of gravity,the torques and end-effector output forces are related by the following equation:

$$\begin{bmatrix} \tau_1 \\ \tau_2 \\ \tau_3 \end{bmatrix} = J^T \begin{bmatrix} f_x \\ f_y \\ n_z \end{bmatrix} \tag{2.34}$$

where

$$J=\begin{bmatrix} -(a_1 \sin\theta_1 + a_2 \sin\theta_{12} + a_3 \sin\theta_{123}) & -(a_2 \sin\theta_{12} + a_3 \sin\theta_{123}) & -a_3 \sin\theta_{123} \\ (a_1 \cos\theta_1 + a_2 \cos\theta_{12} + a_3 \cos\theta_{123}) & (a_2 \cos\theta_{12} + a_3 \cos\theta_{123}) & a_3 \cos\theta_{123} \\ 1 & 1 & 1 \end{bmatrix}$$

Hence, in the absence of gravity, the transformation between the end-effector output forces and the joint torques is governed by the transpose of the conventional Jacobian matrix.

CHAPTER THREE

TYPES OF AMPUTATION THAT PREVENT WALKING

3.1 Limb Amputation

There are several levels at which the surgeon can amputate a limb that is shown Figure 3.1. The most common are:

• Through the foot (transmetatarsal)
• Ankle (ankle disarticulation – Syme)
• Below the knee (transtibial)
• Through the knee (knee disarticulation – Gritti Stokes)
• Above the knee (transfemoral)

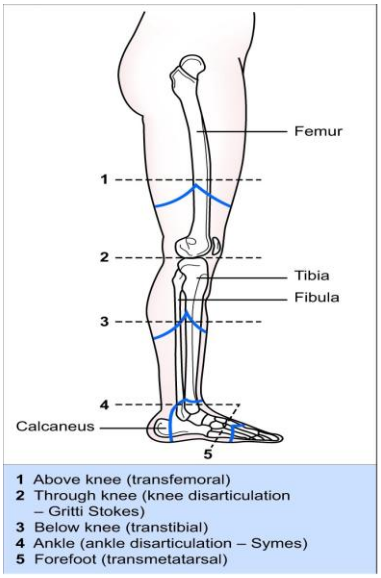

Figure 3.1 Common levels of limb amputation

The level of amputation depends on where there is the greatest blood flow and, therefore, the greatest possibility of healing. The surgeon often attempts to save the knee, because the energy cost of walking with an intact knee is much less than without it. The most common problems during the immediate post-surgery period are wound healing, infections, limited range of motion, swelling, and pain in the residual limb. The goals of this part of recovery are adequate healing of residual limb, optimizing nutrition, minimizing pain and swelling, and slowly starting the rehabilitation process. (Cristian, 2005)

3.2 Syme Amputation

A Syme amputation was named for James syme, a noted University of Edinburg surgeon, in the mid-1800s. This amputation is an ankle disarticulation in which the heel pad is kept for good weight bearing. The Syme amputation results in a residual limb that possesses good function due to the long lever arm to control the prosthesis and the ability to ambulate without the prosthesis.

Associated problems with the Syme amputation include an unstable heel flap, development of neuromas of the posterior tibial nerve, and poor cosmesis. Performed properly, the residual limb is ideally suited for weight bearing and lasts virtually the life of the patient.

The bulky residual limb that results from a Syme amputation may be streamlined by trimming the remaining metaphyseal flares of the tibia and fibula.

3.3 Transtibial Amputation

Transtibial amputation levels are divided in three parts.

3.3.1 A Very Short Transtibial

A very short transtibial amputation occurs when less than 20% of tibial length is present. This amputation may result from trauma and is usually not done as an elective procedure. A very short transtibial amputation results in a small-moment arm, making knee extension difficult.

3.3.2 A Standard Transtibial

A standart transtibial amputation occurs when between 20 and 50% of tibial length is present. An elective amputation in the middle third of the tibia, regardless of measured length, provides a well-padded and biomechanically sufficient lever arm.

At least 8 cm of tibia is required below the knee joint for optimal fitting of a prosthesis.

3.3.3 A Long Transtibial

A long transtibial amputation occurs when more than 50% of tibital length is present. This amputation is not advised because of poor blood supply in the distal leg.

The level of tibial transaction should be as long as possible between the tibial tubercle and the junction of the middle and distal thirds of the tibia. A long posterior flap for transtibial amputations are advantageous because it is well vascularized and provides an excellent weight-bearing surface. In addition, the scar is on the anterior border, an area that is subject to less weight bearing. The deep calf musculature is often thinned to reduce the bulk of the posterior flap.

In a transtibial amputation, the fibula is transected 1 to 2 cm shorter than the tibia to avoid distal fibula pain. If the fibula is transected at the same length as the tibia, the patient senses that the fibula is too long which may cause pain over the distal fibula. The fibula is cut too short, a more conical shape, rather than the desired cylindrical – shape residual limb results. The cylindrical shape is better suited for total contact prosthetic fitting techniques. A bevel is placed on the anterior distal tibia to minimize tibial pain on weight bearing. To avoid a painful neuroma, a collection of axons and fibrous tissue, nerves should be identified, drawn down, severed, and allowed to retract at least 3 to 5 cm away from the areas of weight-bearing pressure.

3.4 Transfemoral Amputation

Still, more transfemoral amputations are required than many people realize. Of the more than 1.2 million people in the United States living with limb loss, 18.5 percent are transfemoral amputees, according to the latest figures provided by the National Center for Health Statistics. The study provided to us by the National Limb Loss Information Center (NLLIC), shows that there were 266,465 transfemoral amputations performed in the United States from 1988 through 1996 (the most recent years available). That's an average of 29,607 annually. Of almost 150,000 amputations performed in the US in 1997, over 35,000 were transfemoral.

Statistically, almost one of every five people living with limb loss in this country has a transfemoral amputation.

Transfemoral amputation is most commonly known as an above-knee amputation, or AK. It's referred to as a transfemoral amputation because the amputation occurs in the thigh, through the femoral bone (femur). Most of these amputations occurred as a result of severe vascular and diabetic disease, with a poor potential to heal a lower level amputation. However, other etiologies included severe soft tissue, vascular, neurological and bone injury resulting from trauma. Additionally, some amputations occurred as a result of severe infection or tumor. Upon amputation, the amputee begins a large rehabilitation process that will involve his surgeon, prosthetist and therapist. But the surgeon has the first and most immediate responsibility, to perform a good amputation. That involves leaving as much residual limb as possible, preserving the adductor, and effective suturing of the remaining soft tissue. It has been shown that the length of the residual limb is inversely related to the energy consumption in walking with prosthesis. Because abduction of the femur is a common problem amongst transfemoral amputees affecting both their gait and energy consumption, preservation of the adductor (to balance the abductor) is important.

While the transfemoral amputation level is fairly common, there's nothing simple about adjusting to life after surgery. The person living with transfemoral limb loss faces distinct challenges, such as increased energy requirements, balance and stability problems, the need for a more complicated prosthetic device, difficulty rising from a seated position, and, unlike with amputation levels in the tibia and the foot, prosthetic comfort while sitting.

3.5 Comparison of Transfemoral Amputation with Others

3.5.1 Energy and Speed

No amputation is “easy” to adapt to, but the transfemoral certainly offers more challenges than amputations in the calf or foot. Figure 3.2 shows that the higher the amputation level, the more energy needed for walking.

A study by Dr. Robert L. Waters and co-workers titled *Energy Cost of Walking of Amputees: The Influence of Level of Amputation*, which was published in *The Journal of Bone and Joint Surgery* (1976), looked at gait and energy use among 70 people with lower-limb amputations. Transfemoral, transtibial and Syme amputations resulting from vascular disease and trauma were compared among the participants with limb loss and to a control group of individuals without amputations. As Graph 1 illustrates, the chosen velocity of walking for vascular amputees was 66 percent of that for nonamputees at the Syme level, 59 percent at the transtibial level and 44 percent at the transfemoral level. Among trauma amputees, velocity was 87 percent for the transtibial level and 63 percent for the transfemoral level. In short, the higher the amputation level, the slower the walking speed. Trauma amputees walked faster than vascular amputees primarily because of age differences and overall health status. By the time blood vessels in the legs are diseased to the point where amputation is needed; individuals with vascular disease also have significant disease of the blood vessels in the heart and lungs. Gait improved and the energy required for prosthetic walking significantly decreased as amputation levels moved toward the foot.

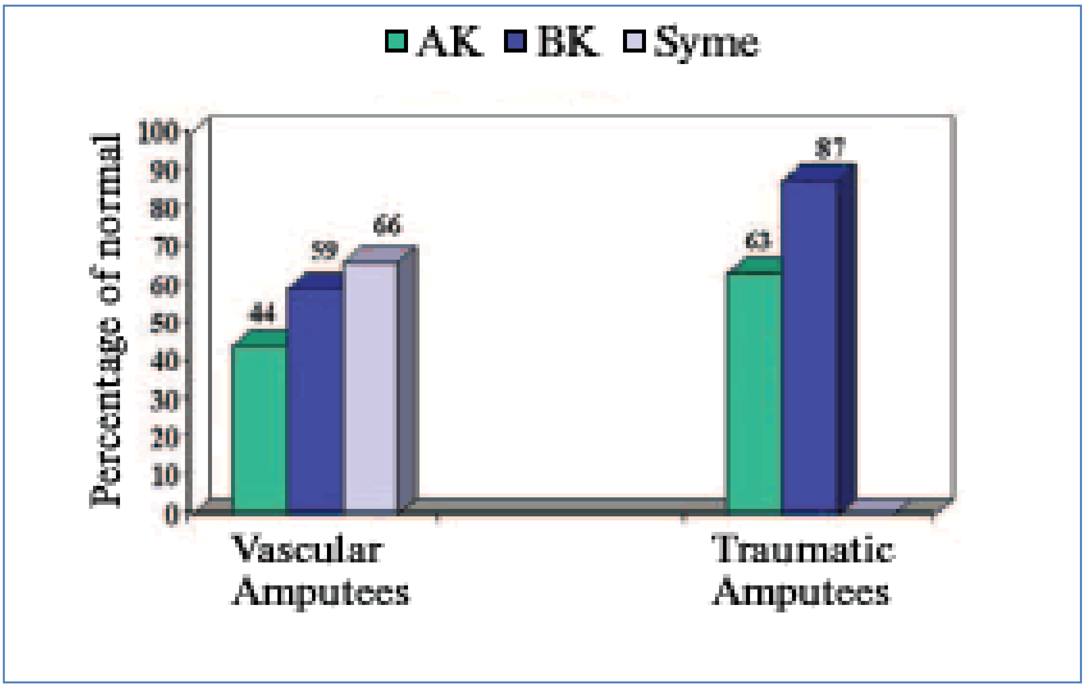

Figure 3.2 Self selected walking velocity

Measuring energy required for walking is tricky. We're not counting just the energy needed for each step; we're also looking at the energy used over a particular distance. In some circumstances, each step for a transfemoral amputee requires more energy than it does for a transtibial amputee, but in other circumstances, the energy per step can be the same or even a little less. Because the stride length for a transfemoral amputee is shorter, however, it takes many more steps to cover the distance. Therefore, when the total energy used by a transfemoral amputee to get from point A to point B is added up, it will probably have taken him or her much more energy than it would have for a transtibial amputee to go the same distance, even though the transfemoral amputee's energy expenditure per step may be less because of the shorter stride.

To measure energy, subjects are outfitted with a mask and a backpack containing an oxygen tank. As the person breathes in and out, sensitive monitoring equipment measures the amount of oxygen being inhaled and exhaled through the mask over a set distance. This oxygen use is then converted to the amount of energy that's required to cover that distance. If your energy requirements increase, you breathe faster and use more oxygen. Figure 3.3 shows that the higher the amputation level, the more energy expended per meter traveled. (Douglas, 2004)

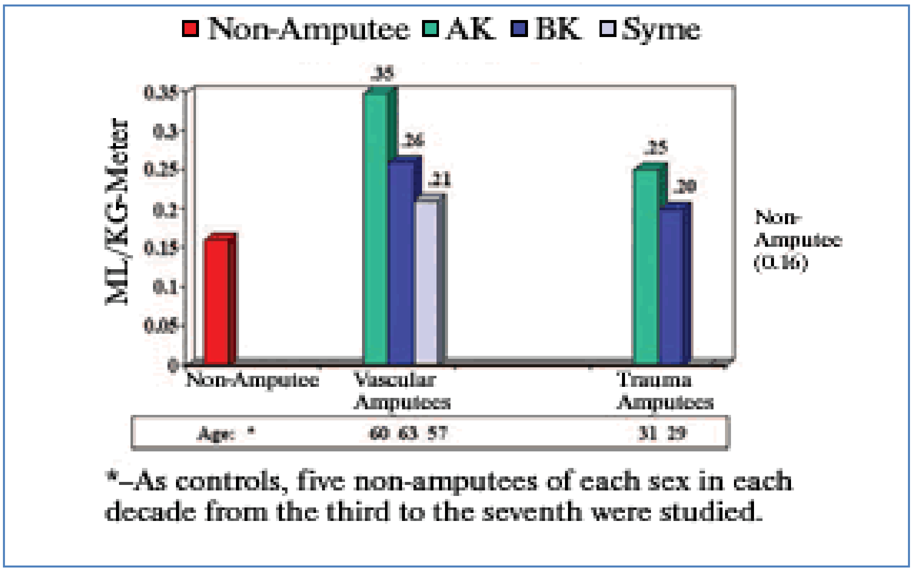

Figure 3.3 Oxygen consumed per meter traveled

CHAPTER FOUR

PROSTHETIC SOLUTIONS PRODUCED FOR AMPUTEES

4.1 History of Prosthetics

The earliest evidence of an amputee is a 45,000-year-old human skull in the Smithsonian Institute that has teeth shaped and aligned in such a way to indicate he was an upper extremity amputee.

4.1.1 The Prosthesis of The Ancient

Cultures began as simple crutches or wooden and leather cups. This evolved into a type of modified crutch or peg to free the hands for everyday functions. An open socket peg leg held cloth rags to soften the distal tibia and fibula and allow a wide range of motion.

With the birth of the great civilizations of Egypt, Greece and Rome came the development of the scientific approach toward medicine and subsequently prosthetic science. Pliny the Elder wrote of Marcus Sergius, a Roman general who sustained injuries and a right arm amputation during the second Punic war (218 and 210 BC).

An iron hand was fashioned to hold his shield, and he returned to battle. The Dark Ages was a time in which there was little scientific illumination. There were not very many prosthetic alternatives available to the amputee except basic peg legs and hand hooks, which only the rich could afford. Knights had cumbersome prosthesis made by their armorers for use in battle, but they were more cosmetic than functional.

4.1.2 The Prosthesis from Renaissance to Today

According to development at medical, science and philosophy, prosthesis during Renaissance were generally made of iron, steel, copper and wood.

In late 1500s, French Army barber/surgeon Ambroise Paré is considered by many to be the father of modern amputation surgery and prosthetic design. He introduced modern amputation procedures (1529) to the medical community and made prosthesis (1536) for upper- and lower-extremity amputees. He also invented an above-knee device that was a kneeling peg leg and foot prosthesis that had a fixed position, adjustable harness, knee lock control and other engineering features that are used in today's devices. His work showed the first true understanding of how prosthesis should function. A colleague of Paré's, Lorrain, a French locksmith, offered one of the most important contributions to the field when he used leather, paper and glue in place of heavy iron in making prosthesis.

In 1696, Pieter Verduyn developed the first nonlocking below-knee (BK) prosthesis, which would later become the blueprint for current joint and corset devices.

In 1800, a Londoner, James Potts, designed a prosthesis made of a wooden shank and socket, a steel knee joint and an articulated foot that was controlled by catgut tendons from the knee to the ankle. It would become known as the “Anglesey Leg” after the Marquess of Anglesey, who lost his leg in the Battle of Waterloo and wore the leg. William Selpho would later bring the leg to the U.S. in 1839 where it became known as the “Selpho Leg.”

In 1843, Sir James Syme discovered a new method of ankle amputation that did not involve amputating at the thigh. This was welcome among the amputee community because it meant that there was a possibility of walking again with foot prosthesis versus leg prosthesis.

In 1846, Benjamin Palmer saw no reason for leg amputees to have unsightly gaps between various components and improved upon the Selpho leg by adding an anterior spring, smooth appearance, and concealed tendons to simulate naturallooking movement. (Figure 4.1)

Figure 4.1 At the Crystal Palace in London, 1851

Douglas Bly invented and patented the Doctor Bly's anatomical leg in 1858, which he referred to as “the most complete and successful invention ever attained in artificial limbs.”

In 1863, Dubois Parmlee invented an advanced prosthesis with a suction socket, polycentric knee and multi-articulated foot. Later, Gustav Hermann suggested in 1868 the use of aluminum instead of steel to make artificial limbs lighter and more functional. However, the lighter device would have to wait until 1912, when Marcel Desoutter, a famous English aviator, lost his leg in an airplane accident, and made the first aluminum prosthesis with the help of his brother Charles, an engineer.

In the World War I, prosthetics were further enhanced because of telephones and phone directories. Medical doctors were able to place illustrated ads, creating more customers.

Following World War II, veterans were dissatisfied with the lack of technology in their devices and demanded improvement. The U.S. government brokered a deal with military companies to improve prosthetic function rather than that of weapons. This agreement paved the way to the development and production of modern prosthesis.

Today's devices are much lighter, made of plastic, aluminium and composite materials to provide amputees with the most functional devices.

4.2 Designing Prosthetic Knee

In general, prosthetic limb has regenerative and electronically controlled prosthetic joints. More specifically, it is converting electrical energy to mechanical energy. The electrical energy can be used for assisting with an amputee’s gait cycle or providing power to various other electrical energy consuming devices associated with the amputee.

Lower limb amputations can be divided commonly in two types;

- Below knee (BK)
- Above knee (AK)

A below knee amputation is related to a line through the tibia and fibula of lower leg; with knee joint remaining intact. An above knee amputation, however, is a transfemoral amputation we know; meaning that the knee joint is also removed.

Designing a prosthetic limb for an above knee amputee is a more complicated process than constructing for a below knee amputee. Below knee prosthesis is fitted to the amputee’s residual leg, with amputee’s knee joint. However, there is no natural knee joint for above knee amputee, an above knee prosthesis should be constructed to simulate knee flexion and extension for amputee’s satisfaction to use the prosthesis for normal walking. For this purpose, the flexible joint connection must be constructed and connected to lower leg portion to an upper socket portion which fits to the amputee’s residual leg. A prosthetic knee allows the amputee to freely swing during the extension part of the gait cycle and also during the flexion part of the cycle. Some artificial knee joints cause problems for amputees such as instability at flexion and extension parts, for instance.

Controlling gait cycle of an above knee prosthetic leg, both basic and electronically controlled passive knee joints must be developed. These knees employ devices such as pneumatic and hydraulic cylinders, magnetic particle brakes, and other similar damping mechanisms, to damp energy generated during the gait cycle to control motion of a prosthetic knee. These damping devices also make resistance to bend knee joint for additional stability. These devices must be designed based on amputee’s weight, gait pattern and motion type, among other factors. In case of an electronically controlled passive prosthetic knee, a software enabled microprocessor adjusts the best. Electronic control systems associated with passive prosthetic knee also needs energy source as a battery for their operation.

The need for a highly active prosthetic knee to limit heel rise and terminal impact requires significant energy consumption by the amputee. The faster an amputee walks, the faster the prosthetic knee must move and the more energy is required for prosthetic leg, but unfortunately, most of energy is lost at the end of heel rise, or at terminal impact. If the amputee tries to walk faster, the energy dissipation will increase rapidly with speed.

In an attempt to solve these and other problems associated with known passive knee joints, active prosthetic knee joints have been developed. However, up until these active knee joints have suffered from various deficiencies including, among other things, the lack of accurate control, the lack of an acceptable actuator for imparting energy to the amputee’s gait cycle, and the inability to produce a sufficient power supply for purposed actuators that can also be easily transported. For instance, Hydraulic or pneumatic active prosthetic joints are developed, and so hydraulic or pneumatic pump is designed.

4.2.1 Developing Prosthetic Knee

According to many researches and studies, the amount of energy consumption for amputee is higher than non-amputee. All prosthetic knees are designed to overcome deficiencies like these.

During normal gait cycle, the human knee has been shown to absorb more energy than it expends. This is true both a normal and prosthetic knee joint.

4.2.2 Parts of Prosthetic Leg

The above knee prosthesis consists of five major parts of the system.

- The socket
- The knee
- The foot
- The components
- The alignment

Although, the knee part has been described in detail in the previous sections, it will be mentioned.

4.2.2.1 The Socket

This is the part of the prosthesis is to attach prosthetic leg to body. It's very important for the amputee’s comfort and for the knee unit work. Almost every transfemoral amputee believes that it is the most important aspect of the prosthesis.(Figure 4.2)

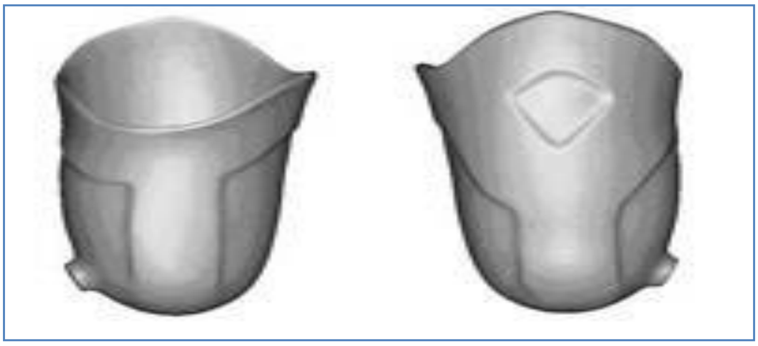

Figure 4.2 Sample sockets for transfemoral amputee

The main functions of socket are to contain and protect the residual limb and to transfer forces from the residual limb to the prosthesis throughout all the amputee's activities (walking, standing, etc.). Therefore, it must enhance comfort and be lightweight. A socket is by definition a Custom-Fit product as it has to be specially manufactured for each patient, following the specific characteristics of the transfemoral limb. It plays a fundamental role for the amputee both in comfort and in the functioning of the prosthesis. That is why Custom-Fit technology is the most appropriate for this type of product. The process to create both, the Check Socket and the Definitive Socket, with CF technology is as follows:

- Definition of amputee's data such as age, weight etc.
- Designing socket according to amputee's characteristic by using CAD program.
- Manufacturing socket

With the improvements in material science; carbon fiber, titanium and graphite sockets are produced especially for transfemoral athletes.

The difficulty in designing an ideal socket was alluded to above: adequate suspension is required during activity such that it does not come loose, as any movement of the socket on the residual limb is likely to lead to skin breakdown and discomfort. Additionally, water loss and muscle contractions throughout a training routine or competition result in continuous changes to the size and shape of a residual limb.

It is only with recent improvements in material science that such challenges can be met, though they are still in need of improvement. Carbon fiber, titanium and graphite are frequently used in socket design. One socket example is that belonging to *John Register,* a Paralympic athlete. He wears a soft flexible plastic socket with a carbon graphite frame. The design contains openings so that his muscles can expand and grow. Too rigid a design can actually lead to muscle atrophy. This socket is a suspension socket, which allows the amputee to be free of belts and straps that might restrict the range of motion. Additionally, the direct skin fit of a soft inner shell of the socket (where the strength is provided by a harder outer frame) provides the amputee with better proprioception and less slipping.

Outside of materials, there are two main designs for sockets, the original quadrilateral socket, and the more recent ischial containment socket. The quadrilateral socket, although known and used by prosthetists for many years, is not ideal for the athlete because it allows lateral socket displacement during stance, compromising pelvic stability. The ischial containment socket, however, is more stable because it locks the ischial tuberosity in the socket. This also helps to stabilize the femur from abducting and improves running gait.

Figure 4.3 Examples of a Quadrilateral socket and an ischial containment socket

On the market, the ComfortFlex Socket from Hanger Orthopaedics is a good choice for athletes. It prevents shifting within the socket by aligning the femur and is contoured with channels and grooves to accommodate muscle, bone, tendon, vascular and nerve areas. This socket also uses a soft flexible inner shell.

Figure 4.4 Example of a ComfortFlex socket

4.2.2.2 The Knee

This is the most important part of the prosthetic leg that involves:

- Knee joint
- Pneumatic cylinder
- Frame
- Microprocessor unit

The main function of knee joint is to support during stance and swing phases. Other functions are to impact absorption during weight acceptance and prevent center of mass rising during the stance phase.

The pneumatic cylinder is to compress air as the knee is flexed, storing energy, and then returning energy as the knee moves into extension.

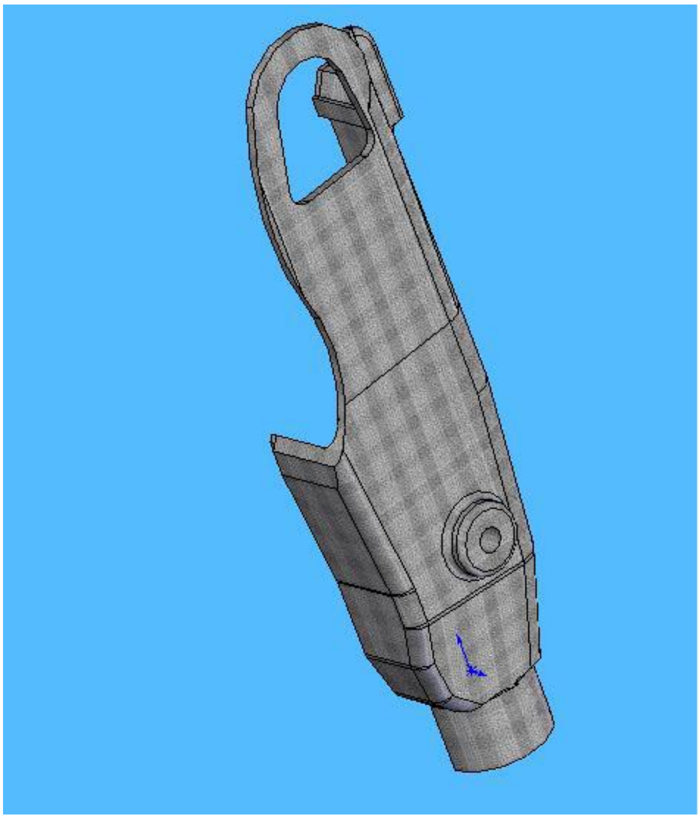

Figure 4.5 Design for prosthesis frame, right side

The knee frame is to cover and protect knee joint, pneumatic cylinder and the microprocessor unit from the environment that is made of carbon fiber composite materials. Carbon fiber composite material is using for lightweight, strength and durability. In figure 4.5, 4.5 and 4.7 is example of knee frame that is designed by using CAD program.

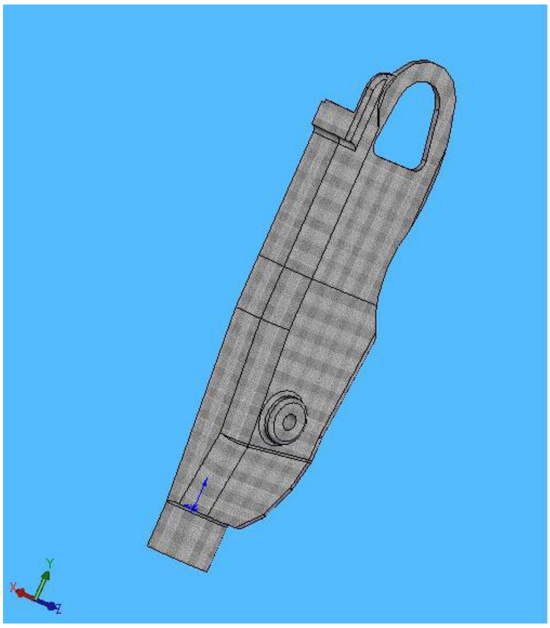

Figure 4.6 Design for prosthesis frame, left side

The microprocessor unit is to control whole knee system. It is initially programmed according to amputee's walking characteristics at various walking situations.

Figure 4.7 Design for prosthesis frame, assembly

4.2.2.3 The Foot

The prosthetic foot is designed according to its tasks such as walking, dancing, cycling, swimming, golfing, snow skiing or running, so fifty models of prosthetic knee are available today. Most of prosthetic knees are made of plastic, metal alloys and carbon-fiber composites to reduce weight and to provide waterproof.

Prosthetic feet can be basic (unmoving), articulated (moving in one or more directions), or dynamic-response (storing and returning energy when walking, giving a sense of "pushing off," much like the human foot). Today's prosthetic feet may have toe and heel springs to allow more ankle movement and adjustable heel heights, and to absorb shock.

There is not only one foot that is perfect for every amputee. The doctor or prosthetist should choose the best prosthetic foot based on amputee's data, age, weight, foot size, activity level, and job needs.

Figure 4.8 The prosthetic foot

4.2.2.4 The Components

These are the parts that replace the various anatomic structures of the lower limb, such as the knee and foot, which were lost at birth or through amputation. These parts range from simple to very complex and are often what people focus on most.

Improvements in the design of and materials for prosthetic foot, ankle and knee components over the last several decades have been truly amazing, but to really appreciate the advantages of technologically advanced components, the amputee must have a good socket and proper suspension.

4.2.2.5 The Alignment

This is the unique way everything fits together – the way the socket, foot and knee are put together in three-dimensional space. Proper alignment ensures that the person isn't too bowlegged or knock-kneed and that the prosthetic knee doesn't buckle when the person stands. Proper alignment means getting the prosthetic knee under the socket in the right spot and the prosthetic foot uniquely positioned beneath the knee and the socket. Good alignment allows the components to accept and support body weight during the stance phase and to bend fluidly as the prosthesis moves through space during the swing phase. (Douglas, 2004)

4.3 A Functional Classification of Knee Mechanisms

4.3.1 Constant Friction Prosthesis

This design group ("single axis" prosthesis) is the oldest historically and consists of a simple axle connecting the thigh and shank segments. These prosthesis are relatively inexpensive and simple to manufacture. Modern versions, such as that manufactured by Otto Bock, have an adjustable friction cell and spring loaded extension assist to improve swing phase function. (Figure 4.9)

Figure 4.9 Single axis constant friction joints

Constant friction knees are best for level ground walking at constant speed but demand sufficient hip power to prevent the knee from buckling. More athletic amputees find this simple design too restrictive.

Biomechanically the Constant friction prosthesis gait resembles that of a patient with a flail leg (eg polio victim). The requirement in both is to keep the ground reaction line in front of the knee from initial contact through mid stance in order to maintain a stable extended knee joint. This ground reaction line should pass behind the knee in terminal stance to ensure ease of knee flexion. Therefore the optimal setting in constant friction prosthesis maintains the ground reaction line within the above parameters.

If a patient lacks hip power and cannot maintain an extended knee in early stance the prosthesis may be adjusted into “hyperextension", by moving the knee center backwards. However this makes knee flexion more difficult during swing phase. The patient must fully unload the knee in order to flex it and this creates the characteristic delayed and abrupt knee flexion on entering the swing phase.

The Constant Friction prosthesis provides only a single fixed cadence during swing phase and therefore if a patient increases his or her walking speed the heel will rise excessively and prolong the swing phase. This encourages the patient to extend the contralateral stance phase by excessively planar flexing the ankle. In other words he vaults over his prosthesis, not because it is too long, but because it is prolonging swing phase on that side. If this patient tries to run he or she hops off the biological leg as this effect is exaggerated. This was the gait pattern demonstrated by Terry Fox, a now famous amputee who attempted to jog across Canada several years ago.

The final problem with the Constant Friction knee is its tendency to give way on declines and on uneven ground.

4.3.2 Stance Control Prosthesis

This knee prosthesis uses a weight activated braking mechanism which adds resistance to bending during stance only. This consists of a spring loaded brake bushing which binds when loaded during stance but is released during swing. The amount of "friction lock" is adjustable. However the brake tends to wear over time and no such device can support full body weight in extreme flexion. The amputee must also delay knee flexion until the device is fully unloaded during swing and this produces an inefficient gait. The device must be fully unloaded before sitting down. This makes it virtually impossible for a bilateral amputee to use Stance Control prosthesis. Biomechanically this knee type best suits the elderly patient with poor hip control. Despite the need for periodic maintenance the Stance Control prosthesis remains very popular. (Figure 4.10)

Figure 4.10 Single axis constant friction joints with weight activated brake

4.3.3 Polycentric Knees

These complex designs comprise multiple centers of rotation. Many have four pivot points and are referred to as “4 bar linkage" devices. Essentially this consists of paired anterior and posterior, superior and inferior hinges linked together. Mechanically the summation of the potential polycentric rotations will determine an instantaneous center of rotation peculiar to a particular device. The stability in polycentric devices is described in terms of “and stability". Stability is determined by the distance that the instant center of rotation is behind the ground reaction line. The greater the distance the greater the inherent stability of the device during stance, just as for the above two types of device. The distance that the instant center of rotation is above the joint line determines the amount of voluntary control the patient has over the prosthesis and is referred to as the stability.

Most Polycentric Knees have their instant centers of rotation quite proximal and posterior for greater stability. Their stability is inherent in their design and not dependent on a brake bushing like the Stance Control device discussed above.

The instant center of rotation moves forward quickly in the swing phase, thus unlocking the joint and facilitating flexion but still offering excellent stance phase stability which allows load bearing during flexion. The polycentric knees shorten slightly during flexion thus adding additional toe clearance during mid swing.

A specific modification of the polycentric knee is available for the knee disarticulation patient, which has long linkage bars placed below the joint line. This offers cosmetic but not mechanical advantage. (Figure 4.11)

Figure 4.11 Four bar knee

4.3.4 Manual Locking Prosthesis

This device offers ultimate stability but is seldom required and produces an uncosmetic and energy - consuming gait pattern. It is useful for the manual laborer who demands stability in the limb. The remote release cable requires a free hand to release it prior to sitting; bilateral device require both hands. The patient falls into the chair with sudden release of both prosthesis. Manual locking devices are rarely used.(Figure 4.12)

Figure 4.12 Single axis with manual lock

4.3.5 Fluid Controlled Devices

These devices utilize a fluid (silicone oil) or gas filled piston which offers automatic hydraulic or pneumatic cadence control respectively. Fluid filled hydraulic devices are stronger. The device allows the amputee to vary their cadence at will. These devices produce the most normal gait parameters. They are relatively heavy and expensive.

All five device types may be incorporated within prosthesis with a soft skin like covering (Endoskeleton) or may be left “exposed" as an Exoskeleton. The exoskeleton "bionic” look seems to have caught the imagination of the American public at least.

Many of the more recent knee prosthesis designs are hybrids which combine some of the properties of the above groups. Otto Bock, for instance, produce a titanium polycentric device which incorporates a mini hydraulic unit for swing phase control. Blatchford, U.K, have produced “bouncy" knees which control knee flexion during stance. Several "intelligent” knees are now available which incorporate microprocessors. (Figure 4.13) (Cormack, 2000)

Figure 4.13 Fluid controlled knees

4.4 Prosthetic Knee Technologies in the World

There are over 100 different prosthetic knee designs available. Space and other limitations make it impossible to showcase every recently marketed prosthetic knee, but here's a representative samples are discussed.

4.4.1 Otto Bock C-Leg

In 1997, Otto Bock HealthCare introduced the C-Leg, the world’s first fully microprocessor-controlled knee. With most prosthetic knees, users worry about stumbling or falling, and have to keep their prosthetic knee straight with each step. But C-Leg Technology changed all that. This remarkable knee immediately set a new standard for stability and performance against which all other knees are measured.(Figure 4.14)

Figure 4.14 Otto Bock C-Leg

C-Leg allows the user to seamlessly speed up or slow down, take on hills or slopes, recover from stumbles and go down stairs step-over-step. The application of science behind the knee is revolutionary by using microprocessors to control the knee’s hydraulic function. The knee is constantly being fine-tuned to adjust to the user’s movements – anticipating what the user is doing and accommodating every change in real-time.

C-Leg has more independence with the Adapting Swing Phase Dynamics feature. This gives C-Leg users the ability to slightly adjust swing phase for higher or lower dynamics for different activities. It's a simple adjustment with the touch of the remote, and it won't compromise the knee's stability.

It has force sensors in the shin that use heel, toe and axial loading data to determine stance phase stability. A knee angle sensor provides data for control of swing phase, angle, velocity and direction of the moment created by the knee. Sensor technology adapts to movement by measuring angles and moments 50 times per second. The unit transfers information to the hydraulic valve allowing reaction to changing conditions. This mechanism results in an individual's gait. It resembles natural walking on many different types of terrain. The C-leg uses a rechargeable battery that lasts 25 to 30 hours. When the battery drains of power, the knee goes into safety mode.

The C-leg was cleared by the US FDA in July 1999 based on its 510(k) application. In this patent application, Otto Bock stated that the C-leg (3C100) is a microprocessor-controlled knee joint system with hydraulic stance and swing phase control. The company claims that C-leg immediately adapts to different walking speeds and provides knee stability. Further, the company stated that C-leg is recommended for lower limb amputees weighing up to 110 kg (220 pounds) who have a moderate (level 2 or 3, i.e. AADL Functional Levels Prosthetic Lower Limb) functional level. The FDA cleared C-leg based on substantial equivalence to a predicate device that was on the market prior to the enactment of the 1976 Medical Device Amendments to the Food, Drug and Cosmetic Act. As such, Otto Bock was not required to provide efficacy data that would be required for pre-market approval.(Craig, 2003)

4.4.2 Ossur Rheo Knee

Manufacturer Ossur collaborated with the Massachusetts Institute of Technology to produce a knee that automatically learns and adapts to the user's movements and adjusts swing and stance resistance for optimal response and stability without the need for programming. (Figure 4.15)

Figure 4.15 Rheo knee

The Rheo Knee checks the force and angular measurements of the user's gait pattern 1,000 times a second and is able to provide instant support to the user. According to Ossur, the magentorheological (MR) actuator reduces fluid drag present in hydraulic knee control systems, allowing for a more rapid foot-off velocity during pre-swing that allows the pelvis to remain in a more normal position. This means the user can walk longer with less fatigue, and gain increased stability and confidence when walking on ramps, varying terrain, and steps.

The Rheo Knee has aluminum frame and does not need programming. It compiles information about the wearer's movements and programs itself. However, a set-up mode does allow a prosthetic practitioner to fine tune parameters.

A lithium ion battery lasts up to 36 hours in constant use and a power switch allows the user to conserve the battery when it is not in use. Charging time is changing between 3 and 4 hours.

The Rheo Knee is ideal for people of moderate and higher activity levels. It allows for cadence variation and ramp or stair descent. The user's weight should not exceed 198 lbs.

4.4.3 Ossur Mauch Knee

There is single axis hydraulic knee system with swing and stance control. Ossur Mauch knee doesn’t have microprocessor for swing and stance control, and so it doesn’t require any battery. The aluminum frame covers the prosthetic knee. (Figure 4.16)

Figure 4.16 Mauch knee

Mauch knee is ideal for short transfemoral amputations because of the enhanced knee flexion lever. Cosmetically good for long residual or knee disarticulation amputations because of the folding linkage and posterior shin set on flexion. Contraindicated for people of small stature because of the knee dimensions.

Mauch Knee Plus is a heavy duty version of the new Mauch Knee. Built on the same advanced technology, it was developed in order to accommodate the needs of users requiring a higher weight limit. A slightly different and more robust frame means that the Mauch Knee Plus is rated up to 166 kg.

4.4.4 Nabtesco Intelligent Knee

Nabtesco intelligent knee has microprocessor controlled knee joint for active patient. The weight limit for the amputee is 100 kg. Nabtesco produced two different types of joints for intelligent knee. (Figure 4.17)

Figure 4.17 Single axis intelligent knee(left), four bar intelligent knee(right)

4.4.4.1 Single Axis Intelligent Knee

It has a carbon fiber composite frame. The braking block is manufactured in titanium. It is lighter than four bar intelligent knee and just only 965 g.

4.4.4.2 Four Bar Intelligent Knee

It has a carbon fiber composite frame, too. The bushings have been replaced with needle bearings that result in lower friction and a more stable knee joint. Its weight is 1097 g.

4.4.5 Endolite Smart Adaptive

The smart adaptive is newest prosthetic knee of Endolite. It has microprocessor control both stance and swing phases. The Smart Adaptive knee addresses mobility in the context of everyday life. It adapts simply to the most complex terrain. The sensors within the system analyze speed, slopes, stairs and other parameters. Adapting to give security, and enhance the amputee’s lifestyle. New Smart programming mode reduces programming time which allows the knee to begin learning the amputee’s gait nuances. (Figure 4.18)

Figure 4.18 Endolite smart adaptive knee

Endolite smart adaptive knee lasts 14 days without recharging. Weight limit for the amputee is 125 kg. Its original weight is approximately 1.3 kg.

4.4.6 Endolite IP Plus

Endolite IP plus has microprocessor control for swing phase and mechanical activated stance phase control. Weight limit for the amputee is 125 kg. Its weight is 1247 g. (Figure 4.19)

Figure 4.19 Endolite IP plus knee

CHAPTER FIVE
AN EXAMPLE ABOUT REMOTE CONTROL DESIGN OF PROSTHETIC KNEE

5.1 Introduction

In this thesis; we tried to design and make a prototype of the remote control of Endolite IP Plus type prosthetic knees. We made some changes but the logic is similar. So that we mentioned some informations about this type of prosthetic knees.

The concept of a microprocessor controlled lower limb prosthesis was first described by Nakagawa et al of Hyaga Assistech in 1986. In 1990 Blatchford obtained the license for this technology and the fruition of their work was the Endolite Intelligent Prosthesis (IP), the first commercial application of microprocessor technology in alower limb prosthesis.

5.2 Development of the Knee

Improvements in the reliability of the complete prosthesis is achieved by a reduction in the forces generated in the mechanism during walking at high speed. The system is designed to adjust itself automatically for different walking speeds, hence, there is no need for amputes to kick their leg to ensure full extension.

In these type prosthetic knees there is new designed cylinder, which is based on it 160 mm geometry, provides 25% more resistance. Equally by utilising the complete extension stroke of the cylinder and controlling the air flow, the cushion for terminal impact has been significantly improved across the entire walking speed range. Thus all categories of amputee walking at every speed can be accommodated. The cylinder is strong enough to take all extension loads, removing the need for secondary extension stops. The type of motor is unchanged. However the drivers have been improved and, in conjuction with a new microprocessor, it is capable of providing half step change which means the valve adjustment can now be fine-tuned to 12 microns instead of 25 microns on the last versions.

In combination with the design of the valve arrangement the system has an effective range which now provides 46 adjustment steps compared to the original 30. This will mean that a smoother and wider range of resistances can be programmed and allowence made for the effect of the cosmetic cover. This can now be prescribed to a larger number of amputees with precise adjustment to the exact speed ef swing on the finished prosthesis.

The smooth resistance profile combined with increased power, greater cushion and damping resistance and the ability to take extension load, means that a larger number of hydraulic users can now be fitted with these type of prosthetic knees. These amputees can now benefit from a significant reduction in effort as well as being able to change their walking speed without any extra effort. This refined control has until now only been available with an hydraulic cylinder, particularly in slow gait. This provides a sure foot step for those amputees who can now safely walk very slowly for the first time. In addition, by using these types of knees, the amputee can walk faster as well, thus enjoying a larger range of walking speeds.

5.3 Controlling the Knee

In order to control the throttle valve of pneumatic cylinder it’s used apparatus at the previous prosthetic knee. In recent years, with occurrence of intelligent knee we can use remote control easy to program for all types of motion. The programming procedure is simply to select a speed of walking, and then adjust the valve position of the swing phase control by pressing an increase / decrease button. The settings are then saved at the required speed. Repeating this sequence at two other speeds, automatically leads to the generation of valve settings.

Firstly, an audible sound confirms receipt of signal and end of task at remote control. The flashing led’s switch to selection of speed, inviting the user to select one of speeds or reset. After selection of a speed, the amputee is asked to walk at that speed at right distance while the prosthetist can observe the gait. The increase (+) or decrease (-) of resistance to flexion controls how fast or slow the limb should be extending. An audible sound confirms each resistance change with the additional feature. Once satisfied with the swing phase performance on any step, the SAVE button is pressed. This stores the selected valve settings as well as the average speed at the time of pressing the key. The sequence is repeated for another two speed selections and this will complete the programming procedure.

It is useful to note that the system goes to automatic mode whenever the SAVE button is pressed. This means that the valve position automatically changed with speed. It is possible to go back to the program and simply adjust the valve setting at one speed or change the walking speed selection at a particular valve setting.

The values stored in permanent memory can only be over written by a new programming sequence.

5.4 Components of Electronic Curcuit

This electronic consist of following components:

- Step motor to control valve position
- 16f628 microprocessor
- ULN2003
- Buzzer for an audible sound
- Led
- Buttons for sequences
- 7805 voltage regulator

It’s designed and simulated using Proteus.

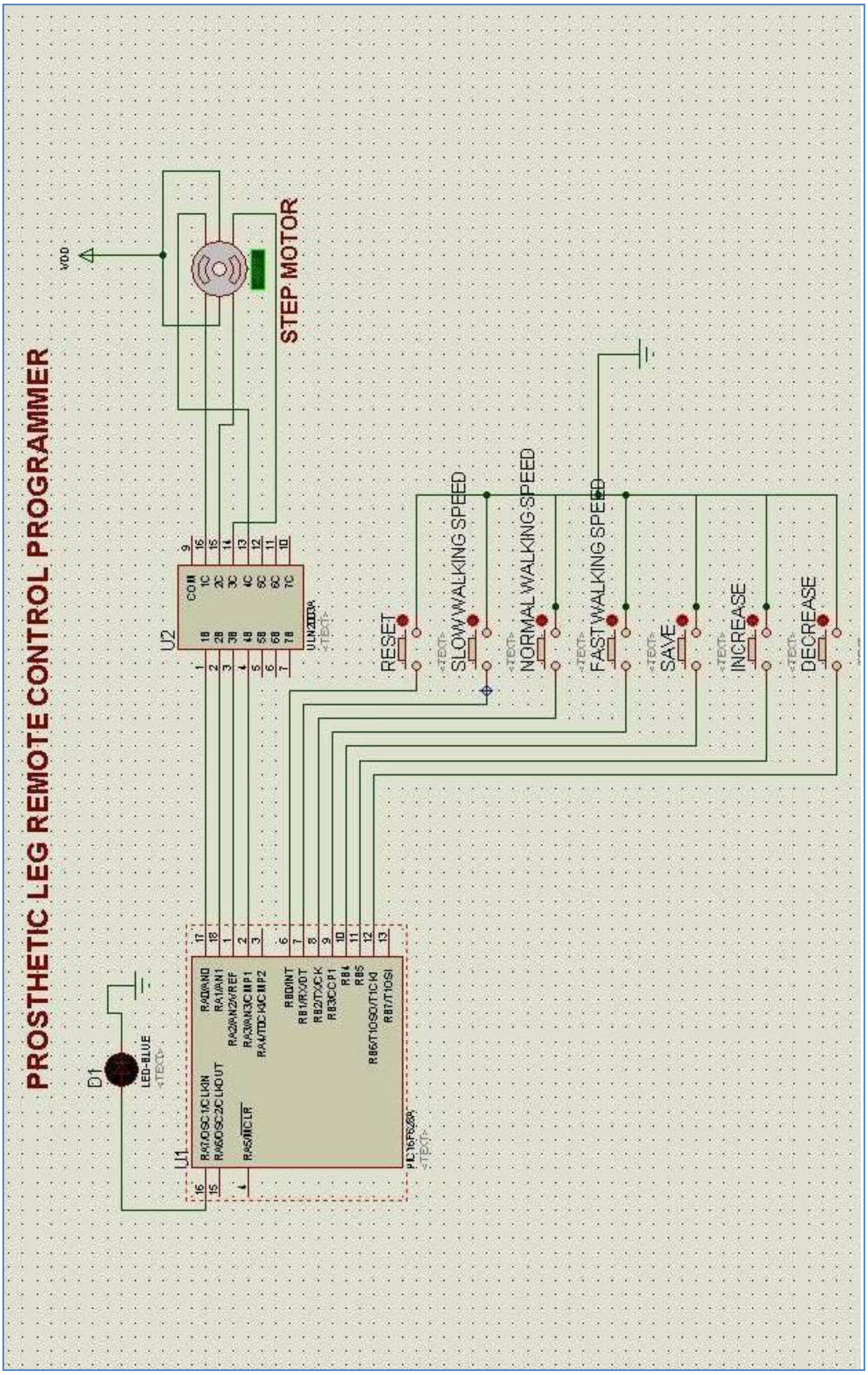

Figure 5.1 Electronic circuit drawing by Proteus

5.5 Programming

The pic programming has been written using micro code studio program in basic language. Then hex file which is being programmed to processor was created by pic basic pro complier. The processor was programmed by using dm programmer with icprog.

Unnecessary components were avoided while the electronic circuit is designing. Screwed step motor is used to control the throttle valve from Copal Electronic Company.

16f628 has been selected to fulfill the tasks of microprocessor control unit from Microchip Company. These properties are:

- High programming and Eeprom memory
- Self contained internal oscillator
- Internal pull up resistances

Having self contained internal oscillator and pull up resistances provided the great simplicity at the control unit electronic circuit.

ULN2003 is integrated to drive unipolar step motors. It has provided simplicity to circuit. Lm7805 voltage regulator is used for conducting stable electric current to step motor and microprocessor. Buzzer is used to receive voice alerts.

In the appendix, the program codes are given in detailed line by line.

CHAPTER SIX
CONCLUSION

This thesis is about to investigate bipedal human walking system. As a result of this investigation, we can model bipedal human walking with using mechanics of bipedal human walking. We can use this modelling for amputees to walk healthy again with any prosthetic legs.

In this thesis we firstly mentioned about the characteristics of bipedal human walking, like its phases. There are two main phases of walking like swing and stance phases. We explained them and then mentioned about anatomical and mechanical meanings to explain principles of walking. Also we evaluated Denavit-Hertenberg Homogeneus Transformation Matrices to model the mechanics of walking. For this evaluation we thought movement of a human leg similar as a planar 3-dof manipulator's movement. With using these matrices we can calculate the position of foot and also positions of members of a human leg and forces which is needed to move this leg.

Modelling of bipedal human walking is important to walk amputees as an healthy person again. So that we must know the types of amputation. The most common levels of amputation are transmetatarsal, Syme, transtibial, Britti Stokes and transfemoral amputation. None of them is easy to adapt to new life for an amputee, but the transfemoral amputation certainly offers more challenges than other amputations in the calf or foot. So that energy requirement of a transfemoral amputee is higher than the others. Since the earliest ages, people tried to find solutions for these amputations. There are many examples of prosthesis down the ages. People used heavy materials during the time but today materials are much lighter; like plastic, aluminium and developing the prosthetic knees of today. Also we must examine the parts of a prosthetic leg and its components. Then we classify prosthetic knees according to their functions. Constant friction prosthesis, stance control prosthesis, polycentric knees, manual locking prosthesis and fluid controlled devices are the types of prosthetic knees according to their functions. Today there are over 100 different knee designs available in the World. Some examples of them:

- Otto Bock C-leg
- Ossur Rheo Knee
- Ossur Mauch Knee
- Nabtesco Intelligent Knee
  - Single Intelligent Knee
  - Four Bar Intelligent Knee Endolite Smart Adaptive
- Endolite Smart Adaptive
- Endolite Intelligent Prosthesis Plus

After these informations about bipedal walking and amputations, in the modelling phase, we gave some informations about an example of prosthetic intelligent knees; then we tried to design and make a prototype of the remote control of Endolite IP Plus type prosthetic knees. We made some changes but the logic is similar. The development of this second generation Intelligent Prosthesis is based on several years experience in the development and commercialisation of the first microprocessor swing phase control. These prosthesis provides functional advantage of adjusting to any speed while reducing the effort in walking. A large number of technical barriers have been overcome in this design, resulting in a user friendly system where the principles of mechatronics in combining electronic controls with mechanical design have been usefully exploited. The rapid adjustment facility, especially when carried out on the finished prosthesis, allowing walking on any terrain outside a clinical environment, has provided the most realistic conditions for normal daily use.

REFERENCES

Amputation and Limb Deficiency, (n.d.) Retrieved February 14,2009, from http://biomed.brown.edu/Courses/BI108/BI108_2003_Groups/Athletic_Prosthetics/Bkgd_text.htm#Glossary

Başer, Ö., (2003). *Analysis and Control of Bipedal Human Locomotion*. Master of Science Thesis. Graduate School of Natural and Applied Sciences, Dokuz Eylül University.

Cormack, M. D. (2000). Prosthetic Knee Designs: Biomechanics and Functional Calssification. *Irish Journal of Orthopaedic Surgery and Trauma 2(1).*

Craig, W.M. (2003). Otto Bock C-leg. Retrieved August 10, 2010 from http://www.worksafebc.com/health_care_providers/Assets/PDF/Otto_Bock_Cleg.pdf

Cristian, A. (2005). *Level of Amputation and Post-Surgical Recovery. Lower Limb Amputation: A Guide to Living a Quality Life* (1st ed.) (2). New York: Demos Medical Publishing.

Douglas, G. S. (2004). The Transfemoral Amputation Level, Part 1 "Doc, It's Ten Times More Difficult!". *In Motion, 14 (2).*

Douglas, G. S. (2004). The Transfemoral Amputation Level, Part 4 Great Prosthetic Components Are Good, But a Good Socket Is Great. *In Motion, 14(5).*

Endolite. (n.d). Retrieved September 18, 2009 from http://www.endolite.com/pdfs/research/Prosthess_Plus.pdf

Highsmith, M. J. & Kahle, J.T. (2005). *Prosthetic Feet.* Retrieved May,2010, from http://www.amputee-coalition.org/military-instep/feet.html

History of Study of Locomotion: Prosthetics. (n.d). Retrieved April, 2010, from http://www.univie.ac.at/cga/history/prosthetics.html

Nabtesco Intelligent Knee. (n.d). Retrieved June, 2010, from http://www.nabtesco.com/en/.

Norton, K. M. (2007). A Brief History of Prosthetics. I*n Motion, 17(7).*

Ossur Rheo Knee. (n.d). Retrieved July, 2010, from http://www.ossur.com/bionictechnology/rheoknee.

Ossur Mauch Knee. (n.d). Retrieved January, 2010, from http://www.ossur.com

Otto Bock C-Leg. (n.d). Retrieved April, 2010, from www.ottobockus.com

Radu, C & Baritz, M.I. (2007). Determination of Normal Cycle Gait Parameters. *Fascicle of Management and Technological Engineering, 4,751-752*.

Tağa, Ö., (2009). *Design and Control of a Lowerknee Prosthesis*. Master of Science Thesis. Graduate School of Natural and Applied Sciences, Dokuz Eylül University.

The History of Prosthetic Devices. (n.d). Retrieved February 14, 2010, from http://www.unc.edu/~mbritt/Prosthetics%20History%20Webpage%20-%20Phys24.html

Transfemoral Prosthesis for the Competitive Athlete.(n.d). Retrieved April 5, 2010, from http://biomed.brown.edu/Courses/BI108/BI108_2003_Groups/Athletic_Prosthetics/AK.htm

Tsai, L.W., (1999). Robot Analysis: The Mechanics of Serial and Parallel Manipulators. New York: John Wiley & Sons Inc.

Zahedi, S., (1996), Senior Research Engineer Of Blatchford Products Division. The Intelligent Prosthesis Plus. Hampshire,The United Kingdom: Endolite Chas. A. Blatchford&Sons Ltd.

APPENDIX

PROGRAMMING CODES

```
 @ DEVICE pic16F628                      'islemci 16F628
 @ DEVICE pic16F628, WDT_OFF             'Watch Dog timer acik
 @ DEVICE pic16F628, PWRT_ON             'Power on timer acik
 @ DEVICE pic16F628, PROTECT_OFF         'Kod Prote
 @ DEVICE pic16F628, MCLR_off 'MCLR pini kullanilMIYOR.
 @ DEVICE pic16F628, INTRC_OSC_NOCLKOUT 'Dahili osilatör kullanilacak
 OPTION_REG.7=0 'dahili Pull up dirençleri deaktif edildi ayrica pullup
direncine gerek yok.
 CMCON=7 '16F628 de komparatör pinleri iptal hepsi giris çikis
 TRISA=%00000000 'A VE B PORTLARININ  GIRIS-ÇIKISLARI AYARLANDI
 TRISB=%11111111
 porta.6=0
 porta.7=0
 say var byte        'degisken boyutu tanimlamalari
 a var byte          'degisken boyutu tanimlamalari
 m var byte          'degisken boyutu tanimlamalari

 c var byte          'degisken boyutu tanimlamalari
 d var byte          'degisken boyutu tanimlamalari

 e var byte          'degisken boyutu tanimlamalari
 f var byte          'degisken boyutu tanimlamalari

 g var byte          'degisken boyutu tanimlamalari
 h var byte          'degisken boyutu tanimlamalari

 read 1,c            'eepromdan 16 bitlik yavas yürüyüs bilgisinin ilk
baytini okur
 read 2,d            'eepromdan 16 bitlik yavas yürüyüs bilgisinin ikinci
baytini okur
 read 3,e            'eepromdan 16 bitlik ortahizli yürüyüs bilgisinin ilk
baytini okur
 read 4,f            'eepromdan 16 bitlik ortahizli yürüyüs bilgisinin
ikinci baytini okur
 read 5,g            'eepromdan 16 bitlik hizli yürüyüs bilgisinin ilk
baytini okur
 read 6,h            'eepromdan 16 bitlik hizli yürüyüs bilgisinin ikinci
baytini okur
 deger var PORTA         'degere degiskenini b portuna yönlendirir.
 i var word              'degisken boyutu tanimlamalari
 l var word              'degisken boyutu tanimlamalari
 lu var word             'degisken boyutu tanimlamalari
 pozisyon var word       'degisken boyutu tanimlamalari
 hizlikonum var word     'degisken boyutu tanimlamalari
 ortakonum var word      'degisken boyutu tanimlamalari
 yavaskonum var word     'degisken boyutu tanimlamalari
 gecicikonum var word    'degisken boyutu tanimlamalari
 hizlikonum=g+h          'eepromdan okunan  8+8bitlik veri birlestirilir.
 ortakonum=e+f           'eepromdan okunan  8+8bitlik veri birlestirilir.
 yavaskonum=c+d          'eepromdan okunan  8+8bitlik veri birlestirilir.
 say=0
```

```
SYMBOL programlama=PORTB.4 'pin sembol ismi olarak programlama adi verildi.
SYMBOL hizliyurume=PORTB.3 'pin sembol ismi olarak hizliyürüme adi verildi.
SYMBOL ortayurume=PORTB.2  'pin sembol ismi olarak ortayürüme adi verildi.
SYMBOL yavasyurume=PORTB.1 'pin sembol ismi olarak yavasyürüme adi verildi.
SYMBOL reset=PORTB.0       'pin sembol ismi olarak reset adi verildi.
SYMBOL yukariadim=PORTB.5  'pin sembol ismi olarak hizliyürüme adi verildi.
SYMBOL asagiadim=PORTB.6   'pin sembol ismi olarak hizliyürüme adi verildi.
SYMBOL LED=PORTA.7       'PortA.0 pinine sembol ismi olarak LED adi verildi.

main :                                  'ana program
if yukariadim=0  then emre              'alt programa yönlendirme
if asagiadim=0  then sura               'alt programa yönlendirme
if reset=0 then resetleme               'alt programa yönlendirme
if programlama=0 then program           'alt programa yönlendirme
if yavasyurume=0 then yavasyuru         'alt programa yönlendirme
if ortayurume=0 then ortayuru           'alt programa yönlendirme
if hizliyurume=0 then hizliyuru         'alt programa yönlendirme
goto main                               'basa dön
end

emre :                            'step motoru ileri dogru döndüren alt program
if say>3 then say=0
lookup say,[ 1,2,4,8],deger       'sinyal tablosu
pause 20                          'adimlar arasi bekleme süresi
porta=deger
say=say+1
pozisyon=pozisyon+1               'pozisyon bilgisi güncellemesi
goto main                         'ana programa dön

sura:                             'step motoru geri dogru döndüren alt program
lookup say,[1,2,4,8],deger
pause 20
porta=deger
if say=0 then say=4
say=say-1
pozisyon=pozisyon-1               'pozisyon bilgisi güncellemesi
goto main                         'ana programa dön

resetleme:     'cihazi resetleyip motoru referans konuma getiren alt program
high led
pause 1000
low led
for i=0 to 550                      'referansa kadar hareketin saglanmasi
if a>3 then a=0
lookup a,[65,66,68,72],deger        'sinyal tablosu
pause 20
porta=deger
a=a+1
next i
pozisyon=0                          'pozisyon bilgisi sifirlanir.
low portA.0 : low portA.1 : low portA.2 : low portA.3
'bir üst satirda step motorun enerjisi kesilir,güç tasarrufu saglanir.
high led
pause 500
low led
goto main                           'ana programa dön
END
```

```
program:                 'hizli ,orta, yavas yürüme konumlari kaydedilir
high led                 'yürüyüs konumu seçilmesini isteyen isik yakilir
if hizliyurume=0 then    'seçilen moda konum verisi kalici olarak kaydedilir
hizlikonum=pozisyon
write 5,0
write 6,0
 if hizlikonum>255 then
 write 5,255
 write 6,hizlikonum-255
 else
 write 5,hizlikonum
 endif
 pause 700
low led

goto main
endif
if ortayurume=0 then
ortakonum=pozisyon
write 3,0
write 4,0
 if ortakonum>255 then
 write 3,255
 write 4,ortakonum-255
 else
 write 3,ortakonum
 endif
 pause 700
low led

goto main
endif
if yavasyurume=0 then
yavaskonum=pozisyon
write 1,0
write 2,0
 if yavaskonum>255 then
 write 1,255
 write 2,yavaskonum-255
 else
 write 1,yavaskonum
 endif
 pause 700
low led

goto main
endif
goto program

hizliyuru:                         'yürüyüs modu alt programi
if hizlikonum>pozisyon then
gecicikonum=hizlikonum-pozisyon    'mevcut pozisyon ve istenen konum
arasindaki fark hesaplanir.
goto stepmotorarti
else
gecicikonum=pozisyon-hizlikonum
goto stepmotoreksi
endif
```

```
ortayuru:
if ortakonum>pozisyon then
gecicikonum=ortakonum-pozisyon
goto stepmotorarti
else
gecicikonum=pozisyon-ortakonum
goto stepmotoreksi
endif


yavasyuru:
if yavaskonum>pozisyon then
gecicikonum=yavaskonum-pozisyon
goto stepmotorarti
else
gecicikonum=pozisyon-yavaskonum
goto stepmotoreksi
endif


stepmotoreksi:    'step motorlari hareket ettirerek valfleri istenen konuma
getirir.
for l=0 to gecicikonum
lookup say,[1,2,4,8],deger
pause 20
porta=deger
if say=0 then say=4
say=say-1
pozisyon=pozisyon-1
next l
goto main

stepmotorarti
for lu=0 to gecicikonum
if say>3 then say=0
lookup say,[1,2,4,8],deger
pause 20
porta=deger
say=say+1
pozisyon=pozisyon+1
next lu
goto main
```