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DESIGN OF A LOWER EXTREMITY EXOSKELETON FOR WALKING POWER AUGMENTATION

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Abstract

Exoskeletons for human performance augmentation (EHPA) are controlled and wearable devices and machines that can increase the speed, strength, and endurance of the operator. Powered exoskeletons can be used to give users certain mechanical advantages.

The objective of this work is to design a control method and develop a lower extremity exoskeleton for walking power augmentation, which can help the user to carry payload and walk with him/her. The system might provide soldiers, fire fighters, disaster relief workers, and other emergency personnel the ability to carry major loads such as food, weaponry, rescue equipment, and communications gear with minimal effort.

In this thesis, a control scheme, which is different from those of the current exoskeleton systems, is introduced. The human leg movements are measured to control the exoskeleton's legs. A special footpad is designed which can measure the ZMP of the human as well as that of the exoskeleton in real time. Using the measured human ZMP as the reference, the exoskeleton's ZMP is controlled by trunk compensation so that the exoskeleton can walk stably.

Secondly, an exoskeleton simulator is developed using ADAMS and MATLAB. The simulator, which consists of adjustable component parameters, is effective and convenient to verify various control algorithms. The simulator can be further used in developing new biped systems.

Design criteria are then set up and function analysis is done to shape the overall concept of the exoskeleton prototype. A few conceptual designs of the prototype are conceived with the help of the morphological chart, and the best conceptual design is selected. Building on this conceptual design, embodiment designs are performed. Subsequently, a detailed final design is achieved and the prototype is constructed.

xPC Target, together with other toolboxes from MATLAB are used so as to provide a

real-time operating system and an integrated development environment. The real-time control is then implemented on the prototype constructed. Experiments show that the control algorithms presented in this thesis are workable.

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Table of Contents

Abstract	i
Acknowledgements	iii
List of Figures	viii
List of Tables	xii
List of Notations	xiii
1 Introduction	1
1.1 Background	1
1.2 Objective and Scope	2
1.3 Contributions of this Research Work	3
1.4 Thesis Layout	4
2 Literature Review	6
2.1 Research Overview	6
2.1.1 Hardiman	6
2.1.2 Exoskeletons for Rehabilitation	7
2.1.3 SpringWalker	10
2.1.4 UC Berkeley's Exoskeletons	11
2.1.5 RoboWalker	14
2.1.6 Hybrid Assistive Leg	14
2.2 Physiology of Human Lower Limb	17
2.2.1 Reference Frames	17
2.2.2 Hip	20
2.2.3 Knee	21

2.2.4	Ankle	22
2.3	Human Gait	23
2.3.1	Terminology Used in Gait Analysis	23
2.3.2	The Walking Process	24
3	Design Considerations	28
3.1	Option of Actuation	28
3.1.1	Electromagnetic Actuators	29
3.1.2	Hydraulic Actuators	29
3.1.3	Pneumatic Actuators	31
3.1.4	Conclusion	32
3.2	Control Scheme and Topology	33
3.2.1	Control Strategies of Exoskeletal Orthosis	33
3.2.2	Objectives of Control	34
3.2.3	Control Scheme	35
3.3	Structure Layout	44
3.3.1	Degrees of Freedom (DoF)	45
3.3.2	Joints and Attachment Points	46
3.4	Safety Issues	47
3.5	Summary	48
4	Control Strategy of the Exoskeleton System	49
4.1	Locomotion Control	49
4.2	Dynamics of the Outer Lower Exoskeleton (OLE)	50
4.3	Control of the ZMP	53
4.4	Measurement of the ZMP	54
4.5	Trunk Compensation	57
4.6	Sensory System of the Footpad	59
4.6.1	Introduction of Selected Sensor	59
4.6.2	Working Principle of the Flexiforce Sensor	60
4.6.3	Sensor Calibration	61
4.6.4	Initial Experiments	63
4.7	Summary	67

5	Simulation	68
5.1	Previous Work	68
5.1.1	Kinematics Approaches	69
5.1.2	Dynamics Approaches	69
5.1.3	Hybrid (Kinematics and Dynamics) Approaches	69
5.1.4	Constrained Optimization	69
5.1.5	Motion Capturing and Editing	70
5.2	Simulation Environment	70
5.3	Human Model Representation	71
5.4	Exoskeleton Model Representation	72
5.5	Walking Simulation	73
5.5.1	Contact Model	73
5.5.2	Locomotion Control	76
5.5.3	Trunk Compensation	76
5.5.4	Simulation Results	80
5.6	Summary	85
6	Design and Construction of Prototypes	86
6.1	Inner Lower Exoskeleton	86
6.1.1	Conceptual Design 1	86
6.1.2	Conceptual Design 2	87
6.1.3	Conceptual Design 3	88
6.1.4	Inner Exoskeleton Harness Design	89
6.2	Outer Lower Exoskeleton	90
6.2.1	Requirements and Constraints	90
6.2.2	Various Conceptual Design	91
6.2.3	Embodiment Design	102
6.2.4	Assembly and Implementation to the OLE	110
6.3	Summary	113
7	Walking Implementation of the Exoskeleton	114
7.1	Real-Time Operating System Used in this Work	114
7.2	Control of the Developed Exoskeleton	118

7.2.1	Joint Control	119
7.2.2	Measurement of the ZMP	124
7.2.3	Exoskeleton's Walking Experiments	127
7.3	Summary	131
8	Conclusions and Recommendations	132
8.1	Conclusions	132
8.2	Future Work	133

List of Figures

2.1	Hardiman	7
2.2	Kinematic walker	7
2.3	“Partial” active exoskeleton	8
2.4	“Complete” exoskeleton	9
2.5	“Complete electrical” exoskeleton	9
2.6	SpringWalker	10
2.7	Two prototypes developed by UC Berkeley	11
2.8	Main components listed in the BLEEX prototype	12
2.9	RoboKnee developed by Yobotics, Inc.	14
2.10	HAL-3 developed by Japan’s Tsukuba University	14
2.11	The composing of HAL-3	16
2.12	The frames of reference used in this thesis	17
2.13	Planes of motion and axes of rotation	18
2.14	Movements about the major joints of the lower limb	19
2.15	Movements of dorsiflexion and plantarflexion of the ankle	23
2.16	Positions of the legs during a single gait cycle	23
2.17	Timing of the single and double support during a single gait cycle	25
2.18	Position of right leg at 40 ms intervals during slightly more than one gait cycle	25
2.19	Angles during a single gait cycle of hip, knee and ankle	26
3.1	Hydraulic human extender	30
3.2	Control setup of the DGO	34
3.3	Definition of ZMP for a kinematic chain	36
3.4	ZMP and CoP	38

3.5	An EMG-based powered exoskeleton arm	40
3.6	An EMG processing system	40
3.7	Human in single support phase	41
3.8	The human and the exoskeleton	42
3.9	Control diagram in swing phase	42
3.10	ZMP trajectory during walking	43
3.11	ZMP trajectory during gait termination	43
3.12	An obstacle inclining the exoskeleton	46
3.13	Structure of the exoskeleton foot	46
3.14	Three passive DoF of the exoskeleton foot	47
4.1	Inner and outer exoskeletons	50
4.2	The model of the exoskeleton	51
4.3	Joint coordinate of a revolute kinematic pair $\{C_{i-1}, C_i\}$	52
4.4	Design of the exoskeleton foot	55
4.5	Sensor distribution	55
4.6	Relationship between the human ZMP and the exoskeleton's ZMP	56
4.7	Adjusting ZMP by trunk compensation	57
4.8	Example of Flexiforce sensor	60
4.9	Excitation circuit	60
4.10	The physical setup of the calibration	61
4.11	Calibration result for Sensor 1 and 2	62
4.12	Generalize equation from Sensor 1 and 2	63
4.13	The test fixture and footpad	63
4.14	Initial experiments	64
4.15	Signals from sensory system when tested on the foot alone	65
4.16	Signals from sensory system when tested with the servomotor	66
4.17	The recorded ZMP trajectory and ground reaction forces acting on the exoskeleton's foot	66
5.1	The human model	71
5.2	The exoskeleton model	73
5.3	Modifying the stiffness	74

5.4	Effect of foot's stiffness on the contact force	75
5.5	Akima splines	77
5.6	The block diagram in Simulink	78
5.7	Torque at the trunk joint	79
5.8	Snapshots of the models while walking	80
5.9	Positions of the legs over a single gait cycle	81
5.10	Ankle flexion/extension torque	82
5.11	Ankle instantaneous mechanical power	83
5.12	Knee flexion/extension torque	83
5.13	Knee instantaneous mechanical power	84
5.14	Hip flexion/extension torque	85
5.15	Hip instantaneous mechanical power	85
6.1	Conceptual design 1	86
6.2	Conceptual design 2	87
6.3	The encoder bracket	88
6.4	The assembled encoder housings	88
6.5	The user wearing the inner exoskeleton	89
6.6	Solutions to attach user at waist	91
6.7	Solutions to adjust length of OLE limbs to user size	94
6.8	Solutions to adjust payload to correct ZMP	96
6.9	Solutions for OLE actuators and transmissions	99
6.10	Morphological chart for the OLE	101
6.11	Final conceptual design	102
6.12	Waist belt	104
6.13	Sliding attachment	104
6.14	Waist assembly	105
6.15	Hip joint	106
6.16	Upper limb	106
6.17	Mounting bracket of the knee joint	107
6.18	Actuated Link of the knee joint	107
6.19	Knee joint	108

6.20	Ankle joint	109
6.21	Ankle joint connected to the foot	109
6.22	The overall design	110
6.23	Photos of the OLE prototype	111
6.24	Photos of the hip joint	111
6.25	Photos of the limb	111
6.26	Photos of the knee joint	112
6.27	Photos of the ankle joint with OLE feet	112
7.1	xPC Target system	115
7.2	Control system architecture of the exoskeleton system	116
7.3	Developed control system set up	117
7.4	Hierarchical Architecture of Control Biped Locomotion Robots	118
7.5	Block scheme of OLE with controller for testing	119
7.6	Response of the left knee joint	120
7.7	Angle difference between the inner exoskeleton and the OLE	121
7.8	Response of the left hip joint	121
7.9	Response of the left ankle joint	122
7.10	Exoskeleton trial with post-stroke patient	122
7.11	Exoskeleton trial with a normal user	123
7.12	Steps of the walking experiment	125
7.13	Ground reaction force below the left foot	125
7.14	Ground reaction force below the right foot	126
7.15	Timing of the ground reaction force below the two feet	126
7.16	ZMP trajectory in the X-Y coordinate system	126
7.17	Signal flow diagram	127
7.18	Block scheme for walking	127
7.19	Human and exoskeleton's ZMP trajectories in the X-Y coordinate system	128
7.20	X coordinate of the human and exoskeleton ZMPs with respect to time	129
7.21	Y coordinate of the human and exoskeleton ZMPs with respect to time	129
7.22	Snapshots of a walking test	130
8.1	Goniometer attached to the knee joint	135

List of Tables

3.1	DoFs of the human lower limbs	45
4.1	Data recorded for sensors 1 and 2	62
5.1	Parameters of the human model’s lower limbs	72
5.2	Mass of the exoskeleton	72

List of Symbols

$\vec{a}_i$	Acceleration of the center of mass of the i -th link
CG	Center of gravity
CM	Center of mass
CoP	Center of pressure
DoF	Degree of freedom
EHPA	Exoskeletons for human performance augmentation
EMG	Electromyogram signal
$\vec{e}_i$	Unit vector of the axis of joint i
$\vec{e}_X$	Unit vector of the X axis of the absolute coordinate frame
$\vec{e}_Y$	Unit vector of the Y axis of the absolute coordinate frame
$\vec{F}_i$	Inertial force of the center of mass of the i -th link
GCP	Ground contact point
G_i	Gravity force of link i
HMCD	Human motion capture data
I_k	Inertia moment for the axis of joint k
$[\vec{i}_i \quad \vec{j}_i \quad \vec{k}_i]$	local orthogonal coordinate frame attached to the center of mass of link i
LEE	Lower extremity exoskeleton
$\vec{M}_F$	Total moment of inertial forces of all the links with respect to ZMP
$\vec{M}_G$	Total movement of gravity forces with respect to ZMP
$\vec{M}_i$	Moment of inertial force of the i -th link
OLE	Outer lower exoskeleton
$\vec{p}^*$	Position vector of joint 1 with respect to the global coordinate system
$\vec{p}_i$	Position vector of the center of mass of the i -th link
$\vec{p}_z$	ZMP coordinates in the global coordinate frame
$\vec{q}$	Vector of generalized coordinates
q_i	Angle of joint i
R_i	Transformation matrix from the i -th link coordinate system into the global system when $q_i \neq 0$
R_i^0	Transformation matrix of the i -th link coordinate frame onto the global frame when $q_i = 0$
$\vec{r}_{ik}$	Position vector from point Z_{ik} to the center of mass of link i

$\vec{v}_i$	Velocity of the center of mass of the i -th link
Z_{ik}	The point at which two adjacent links $\{C_i, C_k\}$ interconnected by a joint
ZMP	Zero moment point
$\vec{\alpha}_i$	Angular acceleration of the i -th link
ΔT_k	Correctional additional actuator torque, applied at joint k
$\Delta \dot{\beta}$	Change in velocity of the upper part
$\Delta \ddot{\beta}$	Change in acceleration of the upper part
$\vec{\omega}_i$	Angular velocity of the i -th link

Chapter 1

Introduction

1.1 Background

Exoskeletons for human performance augmentation (EHPA) are controlled and wearable devices and machines that can increase the speed, strength, and endurance of the operator.

Such devices have long been the stuff of science fiction and movies. In his 1959 novel “Starship Troopers”, Robert Heinlein wrote a story about soldiers of the far future wearing powered suits with missiles and hydrogen bombs to fight alien bugs. A soldier need not drive or operate the suit. He just wears it and it takes orders directly from his muscles.

Remember the fight scene at the end of the movie “Aliens” which was released in 1986? Sigourney Weaver fended off the alien hive queen wearing a machine that looks like a cross between an exoskeleton and a forklift.

There is no doubt that such devices are useful to soldiers. While an army may have laser-guided bombs, aircraft carriers, missiles and so on, now the soldiers still remain to be the key strength of the army because ultimately it wins wars by putting people on the ground and taking territory. Since many years ago, the poor soldiers carrying their equipments — some time up to 150 pounds — have marched on their feet through muddy roads, across hot deserts, up sharp rocks, through raging waters and that wasn’t counting what the enemy was throwing at them at the time that the infantrymen were doing all of this! EHPA can help soldiers avoid such a predicament. It can increase lethality and survivability of ground forces in all combat environments. EHPA is expected to do that by increasing the physical performance of the soldier wearing it, including:

CHAPTER 1. INTRODUCTION

- Increased payload: more fire power, more supplies, and thicker and heavier armor increasing the soldier chance of surviving a direct hit or even an explosion.
- Increased speed and extended range: enhanced ground reconnaissance and battle space dominance.
- Increased strength: larger caliber weapons, obstacle clearance, repairing heavy machinery such as tank.

In the Sept-11 attack, hundreds of fire-fighters and rescue men died when they were trying to rescue people and put out the fire. In the future, terrorists may plant bombs, chemical or biological agents and even nuclear weapons in those big cities. The policemen and fire-fighters will face more dangers than those in Sept-11 attack. The exoskeletons, made of suitable materials, can provide protections against blast, fire, heat, steam, gas, chemical and biological agents, radiation and so on.

In non-military areas, one of the most possible applications of exoskeletons may be used to help aged or disabled people whose lower extremities have locomotor deficiencies due to various reasons: polio, paraparesis, paralysis, dystrophia, etc. They are unable to walk without assistance and may lose muscular strength in their legs and become bedridden. They can only move around by a wheelchair or by using a wheeled walker. Unfortunately, barriers such as bumps and steps restrict the area that these people have access to. It is hoped that the lower exoskeleton can enhance their muscular strength and enable them walk as normal people.

Moreover, if the exoskeleton is simple, compact and cheap enough, even some healthy people will buy one to help them walk or climb stairs. This is more possibly true for those who need walk much everyday, such as postmen, deliverers, etc.

1.2 Objective and Scope

In light of the recent surge of interest in exoskeletons, much research has been devoted to developing exoskeleton systems. However, most of these studies only focus on upper extremity exoskeletons, which are ground-based. Lacking the locomotion capabilities for walking with the user, their application is inevitably limited.

CHAPTER 1. INTRODUCTION

With fund from the Ministry of Defense, Singapore, we are considering to develop a lower extremity exoskeleton (LEE). The LEE aims to help the user carry heavy loads by transferring the load weight to the ground (not to the wearer). It could provide soldiers the ability to carry heavy loads such as food, communications gear, and weaponry, without the strain typically associated with demanding labor. We also hope that with improvements it might provide a versatile transport platform for mission-critical equipment.

The scope of this research is:

- Study and examination of the key technologies required to successfully build an exoskeleton;
- Investigation of the control strategy and the exoskeleton to user interface;
- Design and construction of an experimental prototype;
- Implementation of the control algorithm on the prototype constructed;
- Realization of dynamic walking.

1.3 Contributions of this Research Work

The main purpose of this research is to provide a solution to the important problems of controlling a lower extremity exoskeleton to walk with the user. Studies from various fields, such as robotics, biomechanics, simulation and real-time control are integrated into developing the LEE and the control algorithms.

The contributions of this work, include:

- Investigation of a new control strategy of the lower extremity exoskeleton. Compared to existing methods, controlling the LEE in this study is relatively simpler and more convenient for the user.
- Study and analysis of the dynamics of the LEE.
- Implementation of measurement and control of the Zero Moment Point (ZMP) of the LEE.

CHAPTER 1. INTRODUCTION

- Development of virtual simulators of a human and a LEE. Those simulators enable researchers to perform interactive dynamic simulations, which can verify the control algorithms and provide information for developing or improving the exoskeletons.
- Design and construction of a LEE, which is modular, low cost, and adjustable to suit different users.
- Realization of dynamic walking. The exoskeleton carrying the payload walks with the user and the user can hardly feel the extra burden.

1.4 Thesis Layout

This thesis is organized as follows:

Chapter 1 provides an introduction to exoskeleton. The motivation, scope and contributions of this study are stated.

Chapter 2 gives the background of the work and its study, consisting of the research overview and a brief introduction of the anatomy of the lower limb as well as human gait.

Chapter 3 describes the design considerations, including option of actuation, control scheme, structure layout and safety issues.

Chapter 4 presents the control strategy of the exoskeleton system in detail, including dynamics of the exoskeleton, locomotion control, measurement and control of the ZMP.

Chapter 5 presents the construction of an exoskeleton model by using ADAMS and MATLAB. The control strategy is simulated using the virtual model and the simulation results are demonstrated.

Chapter 6 describes the design and fabrication of the lower extremity exoskeleton. The function of each part of the exoskeleton is introduced and the assembly of the exoskeleton are demonstrated.

CHAPTER 1. INTRODUCTION

Chapter 7 presents the real-time control of the exoskeleton and walking experiments are performed.

Chapter 8 concludes this thesis with an outline of future works.

Chapter 2

Literature Review

In order to understand the work better, the literature review is done in this Chapter. First, previous works on lower extremity exoskeleton systems are identified. What follows is a brief summary of the anatomy of the lower limb as well as human gait since the exoskeleton to be developed is used to help people walk.

2.1 Research Overview

Throughout the last three decades, several designs of exoskeletons, have been developed and evaluated. This section lists various current exoskeleton systems and the various technologies they have used.

2.1.1 Hardiman

The exoskeleton in Figure 2.1 is known as Hardiman, which was built by General Electric with military funding and tested in 1968 [1]. It is a 1,500-pound (≈ 680 kg), 30-DoF, hydraulic and electric full body suit and is controlled as a master-slave follower system¹. But they could only get one arm to work [2]. The project was abandoned for lack of interest by the potential users [3].

¹E. Garcia, Exoskelton for Human Performance Augmentation, Mar. 2000,
http://www.arpa.gov/dso/thrust/md/Exoskeletons/presentations/mar_03_2000/garcia2.pdf.

CHAPTER 2. LITERATURE REVIEW

Figure 2.1: Hardiman

2.1.2 Exoskeletons for Rehabilitation

(a) Kinematic walker

(b) Trials with kinematic walker

Figure 2.2: Kinematic walker [1]

In 1967, the Biocybernetic Department of the “Mihailo Pupin” Institute in Belgrade,

CHAPTER 2. LITERATURE REVIEW

Yugoslavia began to develop exoskeletons for rehabilitation of disabled people. In 1969, a “kinematic walker” as shown in Figure 2.2 was developed. Here, motion of the legs could be only in the sagittal plane², which was not sufficient to satisfy even the nominal conditions of dynamic equilibrium during the gait.

Figure 2.3: “Partial” active exoskeleton [1]

In 1971, the same research team constructed a “partial” exoskeleton (Figure 2.3). Trials proved that a full paraplegic can walk with the help of the exoskeleton only when supported by two people or in a rolling aid. It was not possible to actuate the body trunk of the patient for aiding stability maintenance [4]. For this reason, a new version of the pneumatic active exoskeleton, called the “complete” exoskeleton, was constructed in the same year (Figure 2.4a). It was characterized by a high corselet, enclosing totally the chest of the patient’s body, which enabled controlling the motion of the sagittal and frontal planes. In that way, it realized a much more stable gait enabling some of trial patients to master stable walking with crutches only (Figure 2.4b).

²The definition of each plane can be seen in Figure 2.12.

CHAPTER 2. LITERATURE REVIEW

(a) “Complete” exoskeleton

(b) Patient in “complete” exoskeleton

Figure 2.4: “Complete” exoskeleton [1]

Figure 2.5: “Complete electrical” exoskeleton [1]

CHAPTER 2. LITERATURE REVIEW

The fact that the motion energy had to be supplied from some source of compressed air confined pneumatically driven exoskeletons to closed, mostly clinical, environments. Hence, the research team switched to develop an active exoskeleton with electromechanical drives. In 1974, the “complete electrical” exoskeleton was developed (Figure 2.5). Experiments of the electric exoskeleton demonstrated about 3-4 times better tracking quality compared with the pneumatic ones [5].

2.1.3 SpringWalker

Figure 2.6: SpringWalker

SpringWalker, as shown in Figure 2.6³, is a body amplifier that can enhance the human running gait. Its patent was filed by Applied Motion in 1990. The heart of its technology, described in U.S. patent 5,016,869, combines two ideas for the first time:

- Leg force is increased and body weight distributed by means of a leveraged exoskeleton.

³SpringWalker, Oct. 2002, <http://www.springwalker.com/>

CHAPTER 2. LITERATURE REVIEW

- A single spring is coupled to both legs. This prevents foot-dragging and enables a natural gait for the user.

2.1.4 UC Berkeley's Exoskeletons

In 2000, the Defense Advanced Research Projects Agency (DARPA) began a program spending \$50 million over five years on developing EHAP to give infantry soldiers an extra edge⁴

One of the facilities that received the research funds from DAPRA is located at the UC Berkeley.

(a) Walking trial with a prototype [6]

(b) Electric human powered extender (legs)

Figure 2.7: Two prototypes developed by UC Berkeley

“The legs of an aluminum skeleton hang from Kazerooni’s backpack, its feet bolted to his boots. The lanky metal framework is part of an experimental robot, powered by a chain saw engine, that rides piggyback on Kazerooni, a mechanical engineering professor at the University of California, Berkeley.” This is the description about the walking trial depicted in Figure 2.7(a) [6].

⁴Exoskeletons for Human Performance Augmentation Projects, Dec. 2002,
<http://www.arpa.gov/dso/thrust/md/Exoskeletons>

CHAPTER 2. LITERATURE REVIEW

Figure 2.7(b) shows Electric Human Powered Extender (legs). Each leg has four serial degrees of freedom, and three additional “false” degrees of freedom are used in the interface between the operator and the machine. To avoid the need of a long foot to support torque at the ankle of the machine, there are no actuators below the knee. At the ankle, rotary and linear bearings provide degrees of freedom that are intended to isolate a foot pedal to which the operator’s foot is connected through a bicycle cleat. This cleat further allows for rotary motion of the foot from side to side. A three-axis force sensor is mounted in between the foot pedal and the ankle bearing assembly.

Figure 2.8: Main components listed in the BLEEX prototype

In 2004, Berkeley Lower Extremity Exoskeleton (BLEEX), as shown in Figure 2.8⁵, was first unveiled. The BLEEX is designed to have the same degrees of freedom similar to those of the pilot: three degrees at the ankle and the hip, and one degree at the knee. However, it is hydraulically actuated only at the hips, knees and ankles to allow flexion and extension of the hip joints and knee joints as well as dorsiflexion and plantarflexion of the ankle joints. The other non-actuated degrees of movement are then spring-loaded to a default standing posture. The exoskeleton connects to the user at the foot by strapping onto

⁵Berkeley Robotics Laboratory, Dec. 2004, <http://bleex.me.berkeley.edu/>

CHAPTER 2. LITERATURE REVIEW

the user's boots. A bendable sole allows for bending of the user's toes, and ankle abduction and vertical rotation are allowed for better flexibility. A torso connects the user's back and hips to the exoskeleton legs. A full-body vest is incorporated onto the torso to avoid discomfort or abrasion to the user. The BLEEX employs a high-tech compact Hydraulic Power Unit (HPU). The stand-alone hybrid power source is able to deliver hydraulic power for actuation and electric power for the computers and sensors for long hours. The HPU is the synthesis of a gasoline engine, a three-phase brushless generator and a hydraulic gear pump. The exclusive design by UC Berkeley enables the HPU to regulate the hydraulic pressure and engine speed via an engine throttle and a hydraulic valve. This unique employment of power supply enables the BLEEX to operate more efficiently for a longer period of time!

The control scheme needs no direct measurements from the pilot or the human-machine interface; instead, the controller estimates, based on measurements from the exoskeleton only, how to move so that the wearer feels very little force. The basic principle for the control of BLEEX rests on the notion that the exoskeleton needs to shadow the wearers voluntary and involuntary movements quickly and without delay [7]. This requires a high level of sensitivity in response to all forces and torques on the exoskeleton. However, an exoskeleton with high sensitivity to external forces and torques would respond to other external forces not initiated by its pilot, the pilot would receive motion from the exoskeleton unexpectedly and would have to struggle with it to avoid unwanted movement. The other concern is that systems with high sensitivity to external forces and torques are not robust to variations and therefore the precision of the system performance will be proportional to the precision of the exoskeleton dynamic model. Hence the model accuracy is crucial to exoskeleton control. In conclusion, that maximizing system sensitivity leads to a loss of robustness in the system.

However inventive or state-of-the-art the BLEEX may sound, actual videos of tests done on the BLEEX show certain drawbacks. First of all, the user seemed hampered and unnatural during walking. Also, the backpack looks bulky and makes the whole system rather unbalanced. The costs of building a BLEEX prototype were also not mentioned by the developers. One would suspect that such a complex system, especially with its high-tech HPU, would cost quite considerably.

CHAPTER 2. LITERATURE REVIEW

2.1.5 RoboWalker

Figure 2.9: RoboKnee developed by Yobotics, Inc.

To help people who are suffering from weakness in their lower extremities, Yobotics, Inc., is developing a powered, wearable device called the RoboWalker. The RoboWalker will augment or replace muscular functions of the lower extremities. In about 2001, Jerry E. Pratt *et al.* produced a prototype powered knee orthotic, called the RoboKnee, shown in Figure 2.9⁶. With the computer, amplifiers, and batteries in a backpack, while not too impressive, the RoboKnee did provide for super-human capabilities by allowing the user to perform deep knee bends almost indefinitely.

2.1.6 Hybrid Assistive Leg

Figure 2.10: HAL-3 developed by Japan's Tsukuba University

⁶RoboWalker, Dec. 2003, <http://yobotics.com/robowalker/robowalker.html>

CHAPTER 2. LITERATURE REVIEW

The Hybrid Assistive Leg (HAL), as shown in Figure 2.10⁷, is developed by Sankai lab of Institute of Systems and Engineering Mechanics, University of Tsukuba. The HAL is an exoskeleton-type power assist system to realize the walking aid for aged people or gait-disorder persons.

The composing of HAL-3 is shown in Figure 2.11. At first, the power assist was performed according to the operator's intention by using electromyogram(EMG) as the primary command signal. However, the operator felt quite uncomfortable. To reduce the discomfort, two measures are adopted in HAL-3. In walking-aid mode, Kasaoka *et al.*[8, 9] control the integrated kinematic motion of HAL-3 and the operator corresponding to the particularity of the operator's muscle-skeletal system. In standing-up aid mode, they predict the discomfort by using floor reaction force and feedforward controller, so that the power assist is performed with the quick response.

To detect the electromyogram, two sensor units are attached on the operators skin near the flexor and the extensor driving the targeted joint. A calibration exercise is necessary in order to obtain the conversion coefficients from the measured myoelectric activities to the estimated muscle contraction torque. For the calibration, HAL should output steady torque pattern as a reference torque, and the operator wearing HAL produces torque to compete against the reference torque without producing co-contraction as far as possible. The calibration exercise should be individually performed on the flexor and extensor of the joint. To obtain values of the conversion coefficients, a conventional least-square method is applied. The conversion coefficients apparently depend on the operator's physical condition and the attached location of the sensor unit. Hence, the calibrating motion must be carried out whenever the operator wears HAL [10]. Obviously, the calibrating motion causes much inconvenience to the user.

⁷Powered Suit HAL (Hybrid Assistive Leg), Dec. 2003, <http://sanlab.kz.tsukuba.ac.jp/HAL/index.html>

CHAPTER 2. LITERATURE REVIEW

Figure 2.11: The composing of HAL-3

CHAPTER 2. LITERATURE REVIEW

2.2 Physiology of Human Lower Limb

The large number of degrees of freedom possessed by humans allow a wide range of motion, but for walking not all are needed. Examination of the placement of joints and their functions will be of importance in designing and controlling an exoskeleton.

First the planes of motion and their respective axes of rotation in relation to human movements are introduced; then, the joint movements of lower limb are described and demonstrated.

2.2.1 Reference Frames

Figure 2.12: The frames of reference used in this thesis [11]

When studying the various joints of the body and analyzing their movements, it is helpful to characterize them according to specific planes of motion. There are three specific

CHAPTER 2. LITERATURE REVIEW

planes of motion in which the various joint movements can be classified, as shown in Figure 2.12.

Figure 2.13: Planes of motion and axes of rotation [12]

- The *sagittal* plane is the plane which divides part of the body into right and left portions.
- The *coronal* plane is the plane which divides part of the body into front and back portions.
- The *transverse* plane is the plane which divides part of the body into upper and lower portions.

As movement occurs in a given plane, the joint moves or turns about an axis that is normal to that plane. The axes are named in relation to their orientation (Figure 2.13).

- The *lateral axis* has the same directional orientation as the coronal plane of motion and runs from side to side at right angles with the sagittal plane of motion.

CHAPTER 2. LITERATURE REVIEW

- The *sagittal axis* has the same directional orientation as the sagittal plane of motion and runs from front to back at right angles with the coronal plane of motion.
- The *longitudinal axis* runs straight down through the top of the head and is at right angles with the transverse plane of motion.

Figure 2.14: Movements about the major joints of the lower limb [11]

Accordingly, three possible movements of a joint could take place about those mutually perpendicular axes, although most joints can only make some of these movements. The directions of these motions for the major joints of the legs are shown in Figure 2.14. The possible movements are:

CHAPTER 2. LITERATURE REVIEW

- Flexion and extension — in the ankle these are called plantarflexion and dorsiflexion, respectively.
- Abduction and adduction
- Internal and external rotation

Those movements will be described one by one in what follows.

2.2.2 Hip

The hip is the proximal joint of the lower limb and, being located at its root, it allows the limb to assume any position in space. Hence, it has three axes and three degrees of freedom.

Movements of flexion and extension of the hip

With the knee extended, flexion reaches 90° ; with the knee flexed, flexion can reach up to 120° or higher.

The range of extension is notably less than that of flexion and is limited by the tension of the iliofemoral ligament. Active extension is of lesser range than passive extension. When the knee is in extension, extension of the hip reaches 20° ; when the knee is flexed, extension of the hip reaches 10° ; this follows from the fact that the hamstrings lose some of their efficiency as extensor of the hip because their contraction has largely been utilized in flexing the knee.

Movements of abduction and adduction of the hip

It is theoretically possible to abduct only one hip but in practice abduction at one joint is automatically followed by a similar degree of abduction at the other joint. This becomes obvious after a 30° abduction, when one first clearly notices tilting of the pelvis, as judged from the displacement of the line joining the surface markings of the two posterior iliac spines. If the long axes of the lower limbs are produced they intersect on the line of symmetry of the pelvis. This indicates that in this position each limb has been abducted 15° . When abduction reaches a maximum, the angle between the two lower limbs is 90° . More abduction can be seen to have occurred symmetrically at both joints so that each limb has a maximum of 45° abduction.

CHAPTER 2. LITERATURE REVIEW

Adduction is the movement of the lower limb medially towards the plane of symmetry of the body. As in the position of reference, both limbs are in contact. There is no pure adduction.

There are movements of combined adduction and extension and of combined adduction and flexion at the hip. There are also movements of abduction at one hip combined with abduction at the other hip. Another combined movement which happens to be very common, is illustrated by the position of a person sitting with legs crossed. Adduction is then associated with flexion and external rotation.

In all these combined movements involving adduction the maximal range of adduction is 30°.

Rotational movements of the hip

As the knee is fully extended, rotation occurs only at the hip. When the subject is lying prone, the position of reference is achieved when the leg is at right angles with the thigh and is vertical. From this position, external rotation occurs with a total range of 30° to 40°; Internal rotation occurs with a total range of 60°.

When the subject is in the “straight” position, the total range of rotation is slightly smaller than in the lying position.

2.2.3 Knee

The knee is the intermediate joint of the lower limb. It is mainly a joint with one degree of freedom which allows the end of the limb to be moved towards or away from its root or, in other words, allows the distance between the trunk and the ground to be varied. The knee works essentially by axial compression under the action of gravity.

Movements of flexion and extension of the knee

These are the main movements of the knee and their range is measured from the position of reference established by the following criterion: the axis of the leg is in line with that of the thigh, i.e. seen from the side, the axis of the femur prolongs that of the leg directly. In this position of reference, the lower limb is at its longest.

CHAPTER 2. LITERATURE REVIEW

Extension is defined as the movement of the posterior aspect of the leg away from the posterior surface of the thigh. There is strictly no absolute extension since in the position of reference the limb is maximally extended. It is, however, possible to achieve passive extension (5° to 10°) from the position of reference.

Flexion is the movement of the posterior aspect of the leg towards the posterior aspect of the thigh. Flexion can be absolute, i.e. from the reference position, and relative, i.e. from any position of partial flexion. The range of knee flexion varies according to the position of the hip and according to whether it is active or passive. Active flexion attains a range of 140° if the hip is already flexed and only 120° if the hip is extended. This difference is due to the fact that the hamstrings lose some of their efficiency with extension of the hip. Passive flexion of the knee attains a range of 160° and allows the heel to touch the buttock.

2.2.4 Ankle

The ankle joint is the distal joint of the lower limb. It is a hinge joint and has therefore only one degree of freedom. It controls the movements of the leg relative to the foot, which occur in a sagittal plane. These movements are essential for walking on flat or rough ground. It is subjected to the entire body weight and to the force generated by the dissipation of kinetic energy when the foot rapidly makes contact with the ground during walking, running or jumping.

Movements of dorsiflexion and plantarflexion of the ankle

The position of reference (Figure 2.15) is achieved when the sole of the foot is perpendicular to the axis of the leg (A). From this position, dorsiflexion (B) is the movement which approximates the dorsum of the foot and the anterior surface of the leg. Conversely, plantarflexion (C) is the movement of the dorsum of the foot away from the anterior surface of the leg so that the foot tends to align itself with the leg. In the diagram it is clear that the range of plantarflexion is distinctly greater than that of dorsiflexion. The range of dorsiflexion is from 20° to 30° and that of plantarflexion is from 30° to 50° .

CHAPTER 2. LITERATURE REVIEW

Figure 2.15: Movements of dorsiflexion and plantarflexion of the ankle [13]

2.3 Human Gait

Firstly, some terminology used in describing biped walk will be defined. Then the human walking process will be described.

2.3.1 Terminology Used in Gait Analysis

The term *gait* is defined as a pattern of footsteps at a particular speed, or the manner of walking or running. The *gait cycle* is defined as the time interval between two successive occurrences of one of the repetitive events of walking.

Figure 2.16: Positions of the legs during a single gait cycle [11]

Figure 2.16 shows the positions of the legs during a single gait cycle from right-heel contact to right-heel contact. The gait cycle of human walking can be broken into two

CHAPTER 2. LITERATURE REVIEW

phases: swing phase and stance (support) phase. A leg is in swing phase when it is moving through the air, and the other leg that is in contact with the ground is in stance phase.

Other basic terminology used in this thesis are defined below:

- Center of Mass (CM)⁸ - The point in a system of bodies or an extended body at which the mass of the system may be considered to be concentrated and at which external forces may be considered to be applied.
- Stride length - the distance between two successive placements of the same foot. It consists of two step lengths.
- Step length - the distance by which a foot moves forward in front of the other one.
- Cadence - the number of steps taken in a given time, the usual units being steps per minute.
- Velocity of walking - the distance covered by the whole body in a given time, in a particular direction.
- Static gait - a walking pattern where the CM must be over the supporting base at all time.
- Dynamic gait - a walking pattern where the CM lies outside the supporting base during at least some parts of the walking pattern.

2.3.2 The Walking Process

Figure 2.17 shows the timings of heel contact and toe off for both feet during a single gait cycle. The stance phase usually lasts for about 60 per cent of the cycle, the swing phase for about 40 per cent. There are two double support phases during a cycle and each lasts for about 10 per cent. However, this varies with the speed of walking, the swing phase becoming proportionately longer, and the stance phase and double support phases shorter, as the speed increases [14]. The final disappearance of the double support phase marks the transition from walking to running. Between successive steps in running there is a flight phase, when neither foot is on the ground.

⁸In the presence of a uniform and parallel gravitational field, it is equivalent to the center of gravity (CG)

CHAPTER 2. LITERATURE REVIEW

Figure 2.17: Timing of the single and double support during a single gait cycle [11]

Figure 2.18: Position of right leg at 40 ms intervals during slightly more than one gait cycle [11]

Figure 2.18 shows the successive positions of one leg at 40 ms intervals, measured over slightly more than one gait cycle. Figure 2.19 shows the corresponding angles at the hip, knee and ankle joints, which are taken from a single walk by a 20-year-old healthy female, walking with a cadence of 109 steps/minute, a stride length of 1.45 m and a velocity of 1.32 m/s.

CHAPTER 2. LITERATURE REVIEW

Figure 2.19: Angles during a single gait cycle of hip (flexion positive), knee (flexion positive) and ankle (dorsiflexion positive) [11]

Upper body: The upper body moves forward throughout the gait cycle. Its velocity varies slightly, being highest during the double support phases and lowest in the middle of the stance and swing phases. The trunk twists about a vertical axis, the pelvis and shoulder girdle rotating in opposite directions. Murray [14] found that the average total excursions of 7 degrees for the shoulder girdle and 12 degrees for the pelvis, in adult males walking at free speed. The whole trunk rises and falls twice during the cycle, through a total range of about 50 mm, being lowest during double support, and highest around mid stance or mid swing. An approximation to this vertical motion can be seen in the position of the hip joint marker in Figure 2.18. The trunk also moves from side to side, once in each cycle, being over each leg during its stance phase. The total range of movement is again about 50 mm. The pelvis, as well as twisting about a vertical axis, also tips slightly, both backwards and forwards and from side to side.

Hip: The hip flexes and extends once during the cycle. The limit of flexion is reached about the middle of the swing phase, and the thigh is then kept flexed until the beginning

CHAPTER 2. LITERATURE REVIEW

of the stance phase. The peak extension is reached before the end of the stance phase, after which the hip begins to flex again.

Knee: The knee shows two flexion and two extension peaks during each gait cycle. It is fully extended before heel contact, flexes early in the stance phase (“stance phase flexion”), extends again around mid stance, then starts flexing again, reaching a peak early in the swing phase (“swing phase flexion”). It extends again prior to the next heel contact.

Ankle and foot: The ankle is usually within a few degrees of the neutral position at the time of heel contact, with the heel slightly inverted and the foot slightly supinated. After heel contact, the ankle plantarflexes, bringing the foot down flat on the ground and moving it into pronation. As the stance phase progresses, the tibia moves forward over the foot, and the ankle angle changes from plantarflexed to dorsiflexed. At heel off, the hind foot lifts and the ankle angle again changes, moving back into plantarflexion until toe off. While the fore foot is on the ground and the heel is off it, the heel is inverted and the foot remains supinated. During the swing phase, the ankle moves back into dorsiflexion until the forefoot has cleared the ground, after which the neutral position is adopted prior to the next heel contact.

Chapter 3

Design Considerations

Conceptual design is the initial stage of the exoskeleton development. This is the stage where conceptual solutions are evolved and explored to meet design specification. Further detailed development will base on these conceptual solutions. This chapter lists several conceptual solutions related to the development of the exoskeleton, including actuation option, control scheme and topology, structure layout, and safety issues.

3.1 Option of Actuation

This section will compare various types of actuator technologies subject to their relevance to the exoskeleton project. Below are some critical parameters that should be concerned when one is choosing actuators:

- power/mass ratio
- power/volume ratio
- stress and strain
- steady-state efficiency
- power expended during load holding
- bandwidth or tracking performance
- auxiliary transmission system

CHAPTER 3. DESIGN CONSIDERATIONS

- ease of control

Depending on the type of actuator, some of the critical design parameters will not be applicable or not available due to design specificity. While others have examined these types of actuators in the past, their relevance to the exoskeleton project will be the main focus of this section. Three conventional actuators: electric motors, hydraulic and pneumatic actuators will be discussed. Besides these, there still are some non-conventional actuators such as piezoelectric, shape memory alloys and electroactive polymers. However, the technologies associated with them are still under development and they are not yet used widely out of laboratories. Hence, we pay our attention only to conventional actuators. After comparing them, one kind of actuator which is suitable for this work will be chosen at the end of this section.

3.1.1 Electromagnetic Actuators

The two prototypes shown in Figure 2.7 are driven by electric motors. The electromagnetic actuators allow high bandwidth force control and precise position control. But they produce low torques at high speeds and thus transmissions need to be used. These increase weight and introduce extra inertia and unwanted phenomena such as friction, noise, backlash and torque ripple. Steady-state transmission efficiency is typically 80 to 90 percent, which means that the motors would have to be 10 to 20 percent larger from a power perspective. Depending on the type of transmission system, such as a harmonic to a planetary type of gearhead, the transmission system can increase the overall package volume by 20-50 percent. The power required for holding rated torque at stalled conditions can range from 5 to 10 percent of the power rating of the motor [15]. Basically, this means that while no mechanical power is being expended, significant power is required to merely hold the load. Another problem is that they are stiff and not at all compliant. Small error in position can make a large force error that makes the control task difficult.

3.1.2 Hydraulic Actuators

Figure 3.1 shows an arm extenders which employ hydraulic actuators. It was designed and built at U.C. Berkeley for loading and unloading aircraft. The payload of this power

CHAPTER 3. DESIGN CONSIDERATIONS

extender is 500 lbf ¹.

Figure 3.1: Hydraulic human extender

Hydraulic actuators are fundamentally different from electric motors in that the amount of flow into a hydraulic actuator is limited by the maximum amount of flow possible from the pumping source and any flow restrictions caused by flow control elements such as servo valves. Thermal limitations associated with electric motors do not apply to hydraulic actuators since the hydraulic fluid cools and lubricates the system. Material restrictions are based only on pressure limits. As a rule of thumb, the power-to-mass ratio is about five times that of an electric motor and the power-to-volume ratio is about ten times that of an electric motor for many applications [15].

Unlike electric motors, the force or torque that can be produced can be easily matched to the application simply by changing the effective area for a linear actuator or the effective displacement for a rotary actuator. Therefore, a transmission system is typically not required. Only the friction at the seals and leakage effects limit the steady-state mechanical efficiency. For linear actuators, high, steady-state mechanical efficiencies are possible and are in the upper 90 percent range. Rotary actuators range from 80 to 90 percent. For load-holding applications, zero power is expended for linear actuators with asymmetrical cylinders and a small amount of power is expended in the 1 to 10 percent range because of leakage for symmetrical cylinders and rotary actuators.

One final point associated with hydraulic servo systems, which needs to be noticed, is the noise problem associated with the pumping system. Hydraulic pumps are typically noisy at fairly low frequency, thus making sound-proofing at reasonable packaging volumes

¹Hydraulic Human Powered Extenders, Dec. 2003, <http://www.me.berkeley.edu/hel/hydraulic.htm>

CHAPTER 3. DESIGN CONSIDERATIONS

difficult because of the large acoustical wavelengths.

3.1.3 Pneumatic Actuators

Figure 2.4 shows an exoskeleton powered by pneumatic actuators. Chou and Hannaford [16] reported static and dynamic length-tension testing results and derived a linearized model of these properties for three different McKibben pneumatic artificial muscles and compared the results with human muscle properties to evaluate the suitability of McKibben actuators for human muscle emulation in biologically based robot arms. Tsagarakis *et al.* [17] described a 7-DoF pneumatic muscle actuator powered exoskeleton, and they also presented the conceptual design of the braided pneumatic muscle actuator. Daerden *et al.* [18, 19] introduced pleated pneumatic artificial muscles (PPAMs). PPAMs have adjustable compliance, which is useful for walking and running machines.

Pneumatic actuators are very similar to hydraulic actuators in many respects. However, a few exceptions are especially noteworthy. Air losses are higher because of fluid leakage through seals. Furthermore, air is significantly more compressible than a hydraulic fluid medium. Compressibility of air will significantly lower the natural frequency of a pneumatic control system. Significant energy can be stored in this medium, which can be used to one's advantage, such as mitigating impact loads, or it can be a safety problem because of the possible burst hazard it presents. In comparison, hydraulic fluid has no significant energy storage capability. Also, as a practical limitation, the supply pressure for a pneumatic storage device is limited to 100 to 150 psi as compared to 3,000 to 5,000 psi, for a hydraulic system. Power-to-weight and power-to-volume ratios for pneumatics can vary depending on the type and application. For elastic-type pneumatic actuators, such as the McKibben muscle [20], these ratios can rival hydraulic actuators.

While at first glance pneumatic actuators hold many advantages such as (1) high power to weight and power to volume (for elastic type pneumatic actuators); (2) good impact load mitigation; and (3) energy storage capability, there are some important limitations. While losses are typically higher than hydraulic and safety is always a concern, the main barrier for pneumatic types of actuators is the bandwidth caused by the compressibility of the air. Based on similar assumptions, for a given stroke length, the natural frequency of hydraulics is 5 to 6 times higher than pneumatic actuators. Regardless of the type of feedback scheme,

CHAPTER 3. DESIGN CONSIDERATIONS

since whatever can be applied to a pneumatic servo can be applied to a hydraulic servo, the closedloop bandwidth of a hydraulic system will be 5 to 6 times higher than a pneumatic servo-system for the same stroke length [15].

3.1.4 Conclusion

From the above discussion, it appears that hydraulic and pneumatic actuators have more advantages than electromagnetic ones. Both of them have high power-to-weight and power-to-volume ratios. They do not need transmission. They need no or very little power for load holding. Pneumatic actuators even have inherent and adjustable compliance, which is very useful for walking or running machines. However, neither of them is suitable for this work.

While hydraulic actuators have noise that is harmful to the operator and the low bandwidth of pneumatic ones makes it difficult to drive the exoskeleton walk at a high speed, the main reason why they are not employed is that they need pipes to supply water or compressed air. Those pipes will limit the moving range of the exoskeleton in a close, small space, like those shown in Figure 2.4. Those in Figure 3.1 are even fixed on the ground.

One of main shortages of the electromagnetic actuators is that they have no compliance, which is important to walking machines. However, this problem could be solved in some ways similar to what have been used on biped robots.

Most of the more successful bipedal walking robots employ compliance in one form or another. Waseda's Wabian uses opposing actuators with nonlinear springs in order to vary joint impedance. Honda's P2 and P3 use force/torque sensors on springy feet and vary the joint trajectories in order for the foot to comply with the ground [21]. Yamaguchi *et al.* developed a foot mechanism with shock absorbing material for WL-12RVI biped robot adapting to a horizontally uneven surface [22, 23].

Ideally, a bipedal robot would have perfect torque sources if each of its joint could vary in compliance from being completely limp (zero torque) to being a rigid position source using high gain feedback. Perfect torque sources are impossible to achieve. However, series elastic actuators [24, 25, 26] are a good solution for approximating torque sources at the joints. They are force-controllable actuators. The underlying idea in series elastic actuators is to purposefully place an elastic element (spring) in series with the output of the actuator, after the gear reduction. By measuring the spring deflection, the output force

CHAPTER 3. DESIGN CONSIDERATIONS

can be computed using Hooke's Law. The motor is then served such that the actual output force matches the desired force. The more compliant the elastic element, the more the inertia and stiction of the motor and gear train can be masked, resulting in low output impedance. Series elastic actuators are used in all of the MIT Leg Laboratory's bipedal walking robots.

Yang and Low [27, 28] designed a flexible gear system between the motor and load, by which the force on the gear can be measured and the fuzzy position/force control can then be implemented. The flexible gear system also acts as a buffer to greatly reduce the contact impact magnitude and frequency. Equipped with the flexible gear system in a knee joint, a robot leg could walk smoothly on an unknown terrain.

In conclusion, the exoskeleton to be developed will adopt electromagnetic actuators and employ compliance in some form. The details of how to employ compliance will be discussed in the structure design.

3.2 Control Scheme and Topology

This section is the most important part and the focus of the work. Before the scheme is described, some control strategies used by others will firstly be introduced and the demands of this work will be analyzed. Understanding the demands will help clarify the motivation for the approach which will be discussed later.

3.2.1 Control Strategies of Exoskeletal Orthosis

To do locomotor training of paraplegic patients, Gery Colombo *et al.* [29] developed a Driven Gait Orthosis (DGO), and the control setup is shown in Figure 3.2.

The basic idea is that the therapist presets the parameters of the gait and the exoskeleton will drive the patient's legs move according to the preset trajectories. The step length, the cadence and the walking speed keep constant until the therapist changes the parameters.

"Active suit", is designed and realized by Vukobratović *et al.* [30, 31]. It is a micro-computer controlled active skeletal orthosis powered by servo electric drives, for adding external motion energy to the hip and knee joints of the handicapped.

The "active suit" is operated by a control unit, consisting of several switches and signal

CHAPTER 3. DESIGN CONSIDERATIONS

Figure 3.2: Control setup of the DGO [29]

lamps. The type of gaits, i.e. gait upon level ground, upstairs or downstairs is selected by the three-position mode selector switch. By throwing the two-position stop-start switch to “start”, gait begins, always with the left leg. The user can adjust the gait pace, stride length and turning to left or right, by turning the corresponding button commands.

3.2.2 Objectives of Control

In the two cases mentioned above, the user only walks on a treadmill (in the first case) or in a room (in the second case). The environment is known and simple. The user presets the gait parameters before she/he begins walking, and the exoskeleton will move according to the preset gait until the user changes the parameters, which does not happen frequently. To adjust the parameters, the first case is to key in a computer, and the second one is to push different buttons, both operations need some time. Obviously, they can not change walking patterns arbitrarily.

In this work, the exoskeleton is supposed to work with a soldier in all combat environments, which are usually open, unpredictable and complex. The soldier could meet

CHAPTER 3. DESIGN CONSIDERATIONS

obstacles, steps, slopes, trenches and so on frequently on his way; One minute he needs to walk slowly, and the next minute he should run very fast, thus he needs to adjust his gait: the angles of his joints, the stride length, the cadence, the speed and so on from time to time. In this situation, it is impossible to preset one or several gaits to cover all the demands. However, too many buttons are impractical. Imagine such a scene: a cannonball is coming down, all soldiers have run away except the one with an exoskeleton, who has to choose the correct button among many of them. In addition, the hands of a soldier are busy most of the time, as they should operate a gun or any other weapon. A soldier in such an exoskeleton must stop firing and adjust the parameters manually when he wants to move; however, every common soldier can shoot at the same time he is moving.

From the discussion above, it can be concluded that the ideal exoskeleton should be: (1) the soldier would wear it as an outer skin, rather than operate it; (2) its functions would optimally become an extension of the soldier's natural movements; (3) additional movements are not necessary and what the soldier needs to do is just walk as usual.

3.2.3 Control Scheme

In order to understand the control scheme better, two key concepts of biped walking, the Zero Moment Point (ZMP) and the Center of Pressure (CoP), are firstly introduced. Secondly, the human-machine interfaces that have impact on this work are explored. Lastly, the control strategies in the single support phase and in the double support phase are presented accordingly.

Zero Moment Point (ZMP) and Center of Pressure (CoP)

For a biped robot, all of the biped mechanism joints are powered and directly controllable except for the joint formed by contact of the foot and the ground. Foot behavior cannot be controlled directly; it is controlled in an indirect way by ensuring appropriate dynamics of the mechanism above the foot. To account for this, the concept of zero moment point (ZMP) [32], which is defined as the point on the ground at which the net moment of the inertial and gravity forces has a zero component along the horizontal axes, has been used. The gait is balanced when and only when the ZMP trajectory remains within the support area. In this case, the system dynamics is perfectly balanced by the ground reaction force

CHAPTER 3. DESIGN CONSIDERATIONS

and overturning will not occur. In the single-support phase, the support polygon is identical to the foot surface. In the double support phase, however, the size of the support polygon is defined by the size of the foot surface and by the distance between them (the convex hulls of the two supporting feet). The ZMP concept provides a useful dynamic criterion for the analysis and synthesis of biped locomotion. The ZMP indicates gait balance during the entire gait cycle and provides a quantitative measure for the unbalanced moment about the support foot and for the robustness (balancing margin) of the dynamic gait equilibrium. For the kinematic chain shown in Figure 3.3, the ZMP can be computed using the D'Alembert's Principle [33] which yields the following ZMP condition:

$$(\mathbf{1} - \vec{e}\vec{e}^T) \sum_{i=1}^n (m_i(r_i - P) \times (-\ddot{r}_i + g) - I_i\alpha_i - \omega_i \times I_i\omega_i) = \vec{0} \in R^3 \quad (3.1)$$

where $\mathbf{1}$ is the 3×3 identity matrix, $\vec{e}$ is the unit vector along the Z axis, r_i and P are as defined in Figure 3.3, n is the number of links in the chain, m_i and I_i are respectively, the mass and moment of inertia of link i , ω_i and α_i are, respectively, the angular velocity and angular acceleration of link i , and g is the acceleration due to gravity. The coordinates of

Figure 3.3: Definition of ZMP for a kinematic chain

CHAPTER 3. DESIGN CONSIDERATIONS

the ZMP can be obtained by solving this equation.

Another term is the center of pressure (CoP) [34], also named the ground contact point (GCP) [35], which is commonly used in biped gait analysis based on force platform or pressure mat measurements. CoP or GCP represents the point on the support foot polygon at which the resultant of distributed foot-ground reaction force acts (Figure 3.3). According to their definitions, it is obviously that in the single-support phase considered and for balanced dynamic gait equilibrium, the ZMP coincides with the CoP. However, in the dynamically unbalanced single-support situation that is characterized by a moment about CoP that can not be balanced by the sole reaction forces, as shown in Figure 3.4, the CoP and the ZMP do not coincide. The ZMP location outside the support area (determined by the vector $\mathbf{r}$ in Figure 3.4) provides very useful information for gait balancing. The fact that ZMP is instantaneously on the edge or has left the support polygon indicates the occurrence of an unbalanced moment that cannot be compensated for by foot reaction forces. The distance of ZMP from the foot edge provides the measure for the unbalanced moment that tends to rotate the biped around the supporting foot and, possibly, to cause a fall. When the ZMP approaches critical areas or even abandons the support polygon, it does not mean that the biped is going to collapse. Theoretically, the rotation around the edge of the foot can still be controlled by means of inertial forces generated by movements of the body parts, or stopped by reconfiguring the system, e.g. additional support by the other foot. However, such control requires more effort and is very sensitive to disturbances and parameter uncertainties. Hence, biped dynamic control should prevent the ZMP excursions close to the edges of the supporting polygon.

Many researchers [35, 36, 37, 38, 39, 40, 41, 42] have proposed methods of walking pattern synthesis based on the ZMP. Basically, these investigations first design a desired or referential ZMP trajectory; then, they derive the torso motion to realize the desired ZMP trajectory.

Human-Machine Interface

Throughout the last three decades, several designs of exoskeletons, as human-powered amplifiers, have been developed and evaluated. In studying the evolution of these systems three basic types with a different human machine interface (HMI) seem to emerge: (1) position-position, (2) force-force, and (3) EMG signal.

CHAPTER 3. DESIGN CONSIDERATIONS

Figure 3.4: ZMP and CoP [34]

The first HMI type was used on the Hardiman (Figure 2.1), which was the first attempt to mechanically design a man-amplifying exoskeleton using a hydraulically powered articulated frame worn by an operator. The Hardiman was comprised as a set of overlapping exoskeletons worn by a human operator. The master portion was the inner exoskeleton that was manipulated by the operator providing position command to the slave. The outer exoskeleton consists of a hydraulically actuated slave, which follows all the motions of the master. Position-position control is commonly implemented in systems that have kinematically similar master-slave manipulators.

The second type exoskeletons have the HMI at the dynamics level utilizing the direct contact forces between the human and the machine measured by force sensors as the main command signal to the exoskeleton. The human wears the extender in a way that lumps them together mechanically. The operator is in full physical contact with the exoskeleton throughout its manipulation [43, 44, 45, 46, 47]. Several experimental extender prototypes have been designed and built in order to study control issues.

A common feature in both the first and the second type exoskeletons is that the operator must apply an action, either kinematic-position command or dynamic-contact force command, in order to trigger the exoskeleton response. Obviously, this sequence of events constitutes a source of delay in both systems.

The electrical manifestation of the neuromuscular activation associated with a contracting muscle is called electromyogram (EMG) signal [48]. A new idea that arose in recent years is that setting the HMI at higher levels of the human physiological (neurological)

CHAPTER 3. DESIGN CONSIDERATIONS

system hierarchy allows one to overcome the electro-chemical-mechanical delay, which inherently exists in musculoskeletal system. This inherent time delay refers to the interval between the time when the neural system activates the muscular system and the time when the muscles and the associated soft tissue contract mechanically and generate moments around the joints. During that time interval, the system will gather information regarding the muscle's neural activation level based on processed neuromuscular signals and the joint position and angular velocity. This information is fed into a myoprocessor (muscle model) which in turn predicts the moment that is to be developed by the physiological muscle relative to the joint. The main advantage of establishing the interface at the neuromuscular level is the ability to estimate the forces that will be generated by the muscles before the mechanical contractions actually occur. This information will be fed into the exoskeleton system so that the exoskeleton amplifies the joint moment by a preselected gain factor by the time the physiological muscles contract. As a result, the reaction time of the human/machine system should decrease, resulting in a more natural control of the task [49].

Clancy and Hogan [50, 51, 52] have found during constant force (isotonic) and constant angle (isometric) conditions, the surface EMG recorded from bipolar electrodes can be modeled as a zero-mean correlation-ergodic, random process which obeys Gaussian distribution. Under those conditions the EMG signal can be correlated to the torque developed by the muscles with respect to a joint [53, 54]. Processed EMG signals along with system identification models which noninvasively estimated muscle forces and joint torques have been used as the control input to myoelectrically controlled prostheses [55, 56, 57]. However, when an exoskeleton is used, the isometric and isotonic conditions, which are valid assumption for controlling a prosthetic device using EMG signals alone, do not hold; therefore, the muscle's force cannot be estimated solely on the EMG signals. This is because the angle of the human limb joint, coupled with the exoskeleton joint, is constantly changing during the exoskeleton operation; as a result, the muscles attached to that joint are changing their length and end-point velocities. Therefore, the muscle model (myoprocessor) has to take into account the muscle's length and velocity in addition to the EMG signal, that defines the muscle activation level, for predicting the force that will be developed by the physiological muscle. The one shown in Figure 3.5 is such an exoskeleton, which was built by Rosen, *et al.* [49].

However, unlike the exoskeleton arm that is ground-based, lower exoskeletons should

CHAPTER 3. DESIGN CONSIDERATIONS

Figure 3.5: An EMG-based powered exoskeleton arm [49]

move with the operator. Thus, the information of position of the operator's legs is more important than that of the force of each muscle, which means the position-position control is better than EMG signal control. Moreover, the hardware for EMG signal processing is quite expensive (about 10-60k Singapore dollars) and bulky. Obviously, its volume (Figure 3.6) is a big challenge to build a compact, light exoskeleton. Another disadvantage of EMG is that it varies with different people, even with different states of a same people; thus, it may give wrong commands to the exoskeleton. Due to the reasons stated, EMG signal control will not be employed in this work.

Figure 3.6: An EMG processing system, where each muscle needs one amplifier

CHAPTER 3. DESIGN CONSIDERATIONS

Single Support Phase

Human gait is divided into *single support phase* and *double support phase* [11]. During the single support phase, the swinging foot moves from its last position to its next position. During the double support phase, the CG and ZMP of the human are transferred from the hind foot to the front foot, and the hind foot is ready to leave the ground and attain its swinging phase. Below we will introduce the control strategy in the single support phase.

Figure 3.7: Human in single support phase

In this phase, the human is similar to a ground-based manipulator with an open kinematic chain. In sagittal plane, for the position of the swinging foot, as shown in Figure 3.7, we have

$$P_h = \begin{bmatrix} x_h \\ z_h \end{bmatrix} = f(l_1, l_2, \dots, l_6; q_1, q_2, \dots, q_6) \quad (3.2)$$

where x_h, z_h are the coordinates of the human swinging foot, l_i is the length of link i , and q_i is the trajectory of joint i . For the exoskeleton, similarly, we have

$$P_{ex} = \begin{bmatrix} x_{ex} \\ z_{ex} \end{bmatrix} = f(L_1, L_2, \dots, L_6; Q_1, Q_2, \dots, Q_6) \quad (3.3)$$

where x_{ex}, z_{ex} are the coordinates of the exoskeleton's swinging foot, L_i is the length of link i of the exoskeleton, and Q_i is the trajectory of joint i of the exoskeleton.

CHAPTER 3. DESIGN CONSIDERATIONS

Making $P_{ex} = P_h$, i.e., imposing that the exoskeleton's swinging foot has the same trajectory with that of the human foot, the exoskeleton will have the human step length and height, and will walk with the human on complicated terrain. For example, in Figure 3.7, the human walks over an obstacle, and the exoskeleton will also walk over that obstacle by repeating the same trajectory.

Figure 3.8: The human and the exoskeleton

To do that and make the control as simple as possible, comparing Eqs. (3.2) and (3.3), first, we make the exoskeleton's links parallel to those of the human, and the length of the exoskeleton's leg will be designed to be adjustable. When the user wears the exoskeleton, the length will be adjusted to be the same as the user's leg, i.e. $L_i = l_i$, as shown in Figure 3.8; then, what we should do is just ensure that $Q_i = q_i$. To achieve this goal, the control diagram shown in Figure 3.9 is employed.

Figure 3.9: Control diagram in swing phase

When the swinging foot contacts the ground, the single support phase will end and the double support phase will begin.

CHAPTER 3. DESIGN CONSIDERATIONS

Double Support Phase

In this phase, if the human wants to go on, the ZMP will be transferred from the hind foot to the fore foot, as shown in Figure 3.10. There are force sensors under the human feet and exoskeleton feet to measure their ZMPs. The exoskeleton's ZMP will be controlled to move following the human ZMP.

Figure 3.10: ZMP trajectory during walking

If the human wants to stop, the ZMP will be transferred from the hind foot to a position between the two feet, as shown in Figure 3.11, and the stand phase will begin. Through the force distribution, the exoskeleton's CG will be kept in the support region, until the human starts the next walking.

Figure 3.11: ZMP trajectory during gait termination

CHAPTER 3. DESIGN CONSIDERATIONS

Summary of the Control Scheme

The algorithm includes three sets of feedback information:

- physiological feedback: the operator uses his/her inherent biosensors and receptors (high level feedback – visualization, low level feedback – muscle spindle, tendon organ, joint receptors, etc.)
- force feedback: the force between the human and the exoskeleton structure, the exoskeleton structure and the environment.
- position feedback: the angles of joints of the human and the exoskeleton.

The outer loop is physiological feedback. The human receives information from the environment, determines the gait, then he walks and the exoskeleton walks together with him/her. The operator need not do any additional movements. He just walks naturally as before and the exoskeleton will move with him automatically using the corresponding gait.

While biped control for autonomous robots has been researched for many years, so far no humanoid robot can walk as fast as a normal human in complicated environments. Humans possess naturally developed algorithms with complex and highly specialized control schemes, using higher and lower neural centers, that enable them to perform very complicated tasks such as locomotion while avoiding object collision. In contrast, artificial control algorithms which govern robot dynamics lack the flexibility to perform in a wide range of fuzzy conditions preserving the same quality of performance as humans. Therefore, the exoskeleton integrated with the inherent human control system is supposed to have more mobility than robots.

3.3 Structure Layout

The design of the exoskeleton structure must address the fact that the structure's primary function is to support the payload and provide the mechanical interface with the operator, while at the same time, the function must be transparent to the operator and enable overall system energy efficiency. In essence, the exoskeleton is wrapped around the operator and supports the payload as well as its own load with the objective of minimizing disturbances

CHAPTER 3. DESIGN CONSIDERATIONS

exerted on the operator. In this section, the DoF of the exoskeleton, the number and location of joints, the number and location of attachment points, and measures for safety will be addressed.

3.3.1 Degrees of Freedom (DoF)

In consideration of the DoF of the exoskeleton, the intricacies of human anatomy must be taken into account. The large number of degrees of freedom possessed by humans allow a wide range of motions but for walking not all are needed; the human low limbs have redundant joints. As described in the previous chapter, the primary DoF of human low limbs are shown in Table 3.1.

Table 3.1: DoFs of the human lower limbs

Joint	DoF	Description
Hip	3	Flexion-extension, abduction-adduction, internal-external rotation
Knee	1	Flexion-extension
Ankle	1	Dorsiflexion-plantarflexion

We plan to only employ one actuator at ankle, knee and hip joints respectively, for the flexion-extension in the first stage. This can be supported by two arguments. First, by observing typical human joints' trajectories it can be noted that the motion range is much greater in the sagittal plane than in the coronal and transverse plane [58] and during walking most movements happen in the sagittal plane. Second, adding unnecessary actuators will increase system complexity and weight; however, most of the time, these actuators do not work. Obviously, the exoskeleton can only walk in a line in the first stage of our work. Later, after the exoskeleton succeeds in walking in the sagittal plane, we will add two passive DoF at the hip joint: for abduction-adduction and internal-external rotation, to allow the exoskeleton to adjust its walking direction.

Human can walk on an uneven surface; however, a small obstacle may break the exoskeleton's balance or damage its structure, as shown in Figure 3.12.

To avoid this case, and absorb the shock when the exoskeleton feet contact the ground, the foot structure will be like the one shown in Figure 3.13. The foot will consist of three

CHAPTER 3. DESIGN CONSIDERATIONS

Figure 3.12: An obstacle inclining the exoskeleton

Figure 3.13: Structure of the exoskeleton foot

main parts: (1) an upper plate linked to the exoskeleton ankle; (2) a lower plate that contacts the ground; and (3) elastomeric material installed between the two plates, which should have excellent impact absorption, durability and flexibility.

This structure will add three more passive DoF to the foot: move along the longitudinal axis, and rotate around the sagittal and lateral axes. Hence, the exoskeleton will be able to walk on uneven ground.

3.3.2 Joints and Attachment Points

Human knee joints are not real simple pin joints, but are complex joints with a changing instantaneous center of rotation, with motion characteristics that are extremely variable between individuals. Furthermore, studies of orthotics and principles of interfacing with human tissue indicate that pressures as low as 70 mm of Hg (about 9 kPa) can significantly damage skin and soft tissue in a few hours [15]. Potential pinching of the tendons (e.g., in the back of the upper knee) and the large contraction and deformation of the muscles in the vicinity of the knee makes attachment of the exoskeleton at the knee extremely difficult and undesirable in meeting the objectives of user comfort and non-impeding physical enhancement. Thus, we set the attachment at the foot and waist, which are natural pressure and attachment areas for humans (shoe, belt).

CHAPTER 3. DESIGN CONSIDERATIONS

Figure 3.14: Three passive DoF of the exoskeleton foot

3.4 Safety Issues

Safety will be a prime concern in the eventual deployment of any exoskeleton because of the high bandwidth of an exoskeleton, which could result in significant motion within 150 ms. For example, the required dynamic response could result in 80 movements of the knee in 150 ms or less. The best-case human-reflex response time is 110 to 150 ms [15]. Thus, there is a significant potential for injury without adequate fault detection and response capability. Because of the human reaction times, it is unreasonable to expect that an operator could activate an emergency stop button in the time required to prevent injury in the worst-case scenario. Hence, three independent levels of system safety will be implemented, namely software safety, electrical safety and mechanical safety.

Software safety will be implemented in the real-time controller. Control algorithms will set position, velocity, acceleration and torque limits based on the human anatomy.

Electrical safety is implemented by automatic emergency stop switches. Once the torque, position, velocity or acceleration reaches the preset limits, the amplifiers of the actuators will switch off the motors.

At last, the exoskeleton structure should include built-in mechanical constraints which keep the exoskeleton joint angle within the average human anthropometric boundaries.

CHAPTER 3. DESIGN CONSIDERATIONS

In some bad cases such as when the power of the exoskeleton is finished, or the exoskeleton is damaged, the operator should be capable of escaping from it as soon as possible. Hence, the attachment should be easily and quickly knocked down. The belt in the waist could be designed like a car's seat belts .

3.5 Summary

This chapter presents the design considerations, and conceptual solutions are evolved and explored to meet design specification. Three conventional actuators are discussed and compared; then, suitable actuators are selected. Secondly, several existing control strategies are introduced and demands of this work are analyzed; after that, the proposed control scheme is explained. Lastly, structure layout as well as the safety issues is described.

Chapter 4

Control Strategy of the Exoskeleton System

This chapter presents the control strategy of the exoskeleton system in detail. An important feature of the exoskeleton system, which is also the main difference between exoskeleton and biped robots, is the human participation in the process of control and decision-making. By introducing the human as part of the control system, some intractable tasks for robots such as navigation, path planning, obstacle crossing and gait selection can be easily undertaken by the pilot instead of the robot complex artificial controller and vision system. However, two problems remain for the exoskeleton controller to solve: how to transfer the pilot's intention to the exoskeleton and how to keep the stability of the exoskeleton. Accordingly, the proposed control strategy can be divided into two parts, namely, *Locomotion control* and *ZMP control*.

4.1 Locomotion Control

During the single support phase, the trajectory of the swinging foot determines the gait parameters such as step length, step height, etc. To make sure that the exoskeleton and the wearer can walk together, the trajectory of the exoskeleton's swing foot should trace that of the user in time. To do that, a mechanical linkage named *inner lower exoskeleton* is attached to the human operator, as shown in Figure 4.1. Accordingly, the exoskeleton that is controllable and carries the payloads is named *outer lower exoskeleton (OLE)*.

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

The inner exoskeleton equipped with encoders is to capture the joint information of the pilot. By forward mapping the angles of human joints to the outer exoskeleton, the human operator can control the OLE's legs intuitively.

Figure 4.1: Inner and outer exoskeletons

4.2 Dynamics of the Outer Lower Exoskeleton (OLE)

Before explaining the ZMP control, the dynamics of the OLE is firstly studied and analyzed in this section. As depicted in Figure 4.2, the exoskeleton composed of the trunk, the pelvis, two shanks, two thighs and two feet, will be considered. The trunk carries the payload and the payload can be seen as a part of the trunk. As shown in Figure 4.2, the vertical Z -axis and horizontal X -axis lie in the sagittal plane. By observing typical human joint trajectories it is noted that the motion range is much greater in the sagittal plane than in other planes [58] and during walking most movements happen in the sagittal plane. Hence, at the first stage, only the joints rotating around the Y -axis are actuated and movements in the sagittal plane are studied.

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

Figure 4.2: The model of the exoskeleton

The dynamical equation that describes the movement of a biped, i.e. the exoskeleton, has the following form:

$$M(\vec{q})\ddot{\vec{q}} = f(\vec{q}, \dot{\vec{q}}) + \vec{Q} \quad (4.1)$$

where $\vec{q} = (q_1, \dots, q_7)^T$ is the vector of generalized coordinates, which are the joint angles. The matrix function $M(\vec{q})$ takes into account the mass distribution, and the vector function $f(\vec{q}, \dot{\vec{q}})$ describes the influence of both inertial forces and gravity. The elements of the vector $\vec{Q}$ are generalized forces applied to the system, while the dots denote time derivatives.

The exoskeleton's structure is a set of rigid bodies interconnected by revolute joints. Two adjacent links $\{C_i, C_k\}$ interconnected by a joint at point Z_{ik} is called a kinematic pair. Figure 4.3 shows a kinematic pair whose joint angle is q_i , where $[\vec{i}_i \ \vec{j}_i \ \vec{k}_i]$ is the local orthogonal coordinate frame attached to the center of mass of link i , $\vec{e}_i$ is the unit vector of the axis of joint i , $\vec{r}_{ik}$ is the position vector from point Z_{ik} to the center of mass of link i . The kinematic pairs compose of kinematic chains. Taking the supporting

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

foot as the C_0 , which is connected to the ground, the exoskeleton has two simple chains $\{C_1, C_2, C_3, C_4, C_5, C_6\}$ and $\{C_1, C_2, C_3, C_7\}$, as shown in Figure 4.2b. Without loss of generality, only the first chain is explained here; the second chain can be calculated in a similar way.

Figure 4.3: Joint coordinate of a revolute kinematic pair $\{C_{i-1}, C_i\}$

We set zero joint coordinates q_i to 0 to define the home position of all the links. As for $R_i^0 = [\vec{q}_{i1}^0 \quad \vec{q}_{i2}^0 \quad \vec{q}_{i3}^0]$, which represents the transformation matrix of the i -th link coordinate frame onto the global frame when $q_i = 0$, there exists

$$R_i^0 = [\vec{e}_i^0 \quad \vec{a}_i^0 \quad \vec{e}_i^0 \times \vec{a}_i^0] [\vec{e}_i \quad \vec{a}_i \quad \vec{e}_i \times \vec{a}_i]^T \quad (4.2)$$

where

$$\vec{a}_i^0 = \frac{\vec{e}_i^0 \times (\vec{r}_{i-1,i}^0 \times \vec{e}_i^0)}{|\vec{e}_i^0 \times (\vec{r}_{i-1,i}^0 \times \vec{e}_i^0)|} \quad (4.3)$$

$$\vec{a}_i = -\frac{\vec{e}_i \times (\vec{r}_{ii,i} \times \vec{e}_i)}{|\vec{e}_i \times (\vec{r}_{ii,i} \times \vec{e}_i)|} \quad (4.4)$$

$$\vec{r}_{i,i+1}^0 = R_i^0 \vec{r}_{i,i+1} \quad (4.5)$$

$$\vec{e}_{i+1}^0 = R_i^0 \vec{e}_{i+1} \quad (4.6)$$

where the vectors marked with the tilde $\sim$ represents that the vectors are expressed in the local coordinate frame. By virtue of the recursive computation according to Eqs. (4.2-4.6), R_i^0 can be obtained. Next, applying the theorem of finite rotations (Rodrigue's formula) for the revolute joints, one obtains

$$\vec{q}_{ij} = [\vec{e}_i \times (\vec{q}_{ij}^0 \times \vec{e}_i) \quad \vec{e}_i \times \vec{q}_{ij}^0 \quad (\vec{e}_i \cdot \vec{q}_{ij}^0) \vec{e}_i] \begin{bmatrix} \cos q_i \\ \sin q_i \\ 1 \end{bmatrix}, \quad j = 1, 2, 3 \quad (4.7)$$

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

therefore, the matrix $R_i = [\vec{q}_{i1} \quad \vec{q}_{i2} \quad \vec{q}_{i3}]$ representing the transformation matrix from the i -th link coordinate system to the global system when $q_i \neq 0$, is determined, where $\vec{e}_{i+1} = R_i \vec{e}_{i+1}$ and $\vec{e}_1$ is known. Hence, the position vectors in the global system are obtained as follows:

$$\begin{aligned}\vec{r}_{ii} &= R_i \vec{r}_{ii} \\ \vec{r}_{i,i+1} &= R_i \vec{r}_{i,i+1}\end{aligned}\tag{4.8}$$

Applying the basic theorems of rigid body kinematics, one obtains the following recursive equations:

$$\begin{aligned}\vec{\omega}_i &= \vec{\omega}_{i-1} + \dot{q}_i \vec{e}_i \\ \vec{v}_i &= \vec{v}_{i-1} + \vec{\omega}_{i-1} \times \vec{r}_{i-1,i} + \vec{\omega}_i \times \vec{r}_{ii} \\ \vec{\alpha}_i &= \vec{\alpha}_{i-1} + \dot{q}_i \vec{\omega}_{i-1} \times \vec{e}_i + \ddot{q}_i \vec{e}_i \\ \vec{a}_i &= \vec{a}_{i-1} + \vec{\alpha}_{i-1} \times \vec{r}_{i-1,i} + \vec{\omega}_{i-1} \times (\vec{\omega}_{i-1} \times \vec{r}_{i-1,i}) \\ &\quad + \vec{\alpha}_i \times \vec{r}_{ii} + \vec{\omega}_i \times (\vec{\omega}_i \times \vec{r}_{ii})\end{aligned}\tag{4.9}$$

where $\vec{\omega}_i$, $\vec{\alpha}_i$ are angular velocity and angular acceleration of the i -th link, respectively; $\vec{v}_i$ and $\vec{a}_i$ are velocity and acceleration of the CM of the i -th link, respectively. The inertial force $\vec{F}_i$ of the CM of the i -th link and the moment $\vec{M}_i$ of the i -th link can then be obtained by using the Newton-Euler equations:

$$\begin{aligned}\vec{F}_i &= m_i \vec{a}_i \\ \vec{M}_i &= I_i \vec{\alpha}_i + \vec{\omega}_i \times I_i \vec{\omega}_i\end{aligned}\tag{4.10}$$

with

$$I_i = R_i J_i R_i^T\tag{4.11}$$

where m_i is the mass of link i , I_i is the moment of inertia of link i with respect to the global system, and J_i is the moment of inertia of link i with respect to the i -th link coordinate system.

4.3 Control of the ZMP

From the definition of the ZMP, we have

$$(\vec{M}_G + \vec{M}_F) \cdot \vec{e}_X = 0\tag{4.12}$$

$$(\vec{M}_G + \vec{M}_F) \cdot \vec{e}_Y = 0\tag{4.13}$$

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

where $\vec{M}_G$ is the total movement of gravity forces with respect to ZMP, $\vec{M}_F$ is the total moment of inertial forces of all the links with respect to ZMP, while $\vec{e}_X$ and $\vec{e}_Y$ denote the unit vectors of the X and Y axes of the absolute coordinate frame. Eq. (4.13) can be further replaced with

$$\sum_{i=1}^7 [(\vec{p}_i - \vec{p}_z) \times (G_i + \vec{F}_i) + \vec{M}_i] \cdot \vec{e}_Y = 0 \quad (4.14)$$

where $\vec{p}_z$ is the vector of ZMP coordinates in the global coordinate frame, $\vec{p}_i$ is the position vector of the center of mass of the i -th link,

$$\vec{p}_i = \vec{p}^* + \sum_{j=1}^{i-1} (\vec{r}_{jj} - \vec{r}_{j,j+1}) + \vec{r}_{ii} \quad (4.15)$$

where $G_i = m_i g$ is the gravity force of link i , $\vec{p}^*$ is the position vector of joint 1 in the global coordinate system. Substitution of Eqs. (4.9), (4.10), (4.11), (4.15) into Eq. (4.14) leads to

$$\sum_{i=1}^7 a_i \ddot{q}_i + \sum_{i=1}^7 \sum_{j=1}^7 b_{ij} \dot{q}_i \dot{q}_j + \sum_{i=1}^7 c_i G_i = 0 \quad (4.16)$$

where the coefficients a_i , b_{ij} and c_i are functions of the generalized coordinates q_i . The trajectories of q_1 to q_6 are determined by the signals measured from the pilot's legs, as mentioned before, while q_7 is determined according to Eq. (4.16) in such a way to ensure that the ZMP lies in the support polygon.

Such ZMP is the desired ZMP; however, the actual ZMP may be different from the desired ZMP due to reasons such as disturbance from the environment or error of the actuators. A footpad that can measure the actual ZMP in real time is thus designed.

4.4 Measurement of the ZMP

In a stable gait, during the single support phase, the GCP of the supporting foot is also the ZMP of the whole biped; during the double support phase, the relationship between the ZMP and the GCP is described by

$$X_p = \frac{f_{Lz} X_L + f_{Rz} X_R}{f_{Lz} + f_{Rz}}, \quad Y_p = \frac{f_{Lz} Y_L + f_{Rz} Y_R}{f_{Lz} + f_{Rz}} \quad (4.17)$$

where

ZMP = (X_p, Y_p, Z_p) : ZMP of the whole biped.

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

$GCP_L = (X_L, Y_L, Z_L)$: GCP of the left foot.

$GCP_R = (X_R, Y_R, Z_R)$: GCP of the right foot.

$f_L = (f_{Lx}, f_{Ly}, f_{Lz})$: ground reaction force at GCP_L .

$f_R = (f_{Rx}, f_{Ry}, f_{Rz})$: ground reaction force at GCP_R .

Figure 4.4: Design of the exoskeleton foot

To measure the ZMPs of the pilot and the exoskeleton, a footpad is designed as shown in Figure 4.4. The human foot will be on the upper plate, and the exoskeleton leg will be connected to the middle plate. There are four force sensors between the upper plate and the middle plate, the middle plate and the lower plate, respectively. The sensors are distributed as shown in Figure 4.5.

Figure 4.5: Sensor distribution

During the single support phase, Sensors 1-4 measure the ground reaction force under the human foot, and the ZMP coordinates of the human in the foot local coordinate frame

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

can be calculated according to

$$ZMP_h = \frac{\sum_{i=1}^4 F_i r_i}{\sum_{i=1}^4 F_i} \quad (4.18)$$

where F_i is the force measured by sensor i at a distance r_i from O , as defined in Figure 4.5. Sensors 5-8 measure the ground reaction force under the whole system (the human plus the exoskeleton). Similarly, the ZMP of the whole system can be calculated by

$$ZMP_w = \frac{\sum_{i=5}^8 F_i r_i}{\sum_{i=5}^8 F_i} \quad (4.19)$$

Figure 4.6: Relationship between the human ZMP and the exoskeleton's ZMP

The ZMP of the exoskeleton is in line with the human ZMP and the whole system's ZMP, and its position can be obtained by

$$ZMP_e = \frac{\sum_{i=1}^4 F_i (r_w - r_h)}{\sum_{i=5}^8 F_i - \sum_{i=1}^4 F_i} + r_w \quad (4.20)$$

in which r_h and r_w , are the coordinates of the human ZMP and the ZMP of the whole system, respectively, as shown in Figure 4.6.

During the double support phase, instead of the ZMPs, GCPs of each foot are obtained from Eqs.(4.18)-(4.20). By substituting those GCPs of the human and the exoskeleton into Eq.(4.17), ZMP of the human and that of the exoskeleton can be obtained accordingly.

Using the measured human ZMP as the reference, the desired ZMP of the exoskeleton can be chosen according to the following criteria:

- During the single support phase, the desired ZMP of the exoskeleton should be in the support foot area.

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

- During the double support phase, the human ZMP shifts from the hind foot to the fore foot, the desired ZMP of the exoskeleton could be in the area between the two feet, but not too far from the measured human ZMP.
- At the end of the double support phase, when the human ZMP enters his/her former foot area, the desired ZMP of the exoskeleton should also enter the exoskeleton's former foot area.

4.5 Trunk Compensation

If the actual (measured) ZMP of the exoskeleton differs from the desired ZMP, trunk compensation will be applied to shift the actual ZMP to an appropriate position.

Without loss of generality, the motion in the sagittal plane, during single support phase, is discussed here, and the trunk compensation in the frontal plane or during double support phase is performed in a similar way.

Figure 4.7: Adjusting ZMP by trunk compensation

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

As shown in Figure 4.7, the actual ZMP differs from the desired ZMP in the direction of the X axis by Δx . And the ground reaction force acting on the exoskeleton is F_z , which can be derived from $F_z = \sum_{i=5}^8 F_i - \sum_{i=1}^4 F_i$. An assumption introduced for the purpose of simplicity is that the action at the trunk joint will not cause a change in the motion at any other joint. In other words, the servo systems are supposed to be sufficiently stiff. Under this assumption, the system will behave as if it was composed of two rigid links connected at trunk joint k , as presented in Figure 4.7. The payload and the exoskeleton trunk are considered as an upper part of total mass m and moment of inertia I_k (as $I_k = I_c + ml_1^2$) for the axis of joint k . Point c is the mass center of the upper part, and the distance from c to k is denoted by l_1 . The lower part, representing the sum of all the links below the trunk joint k including the other leg that is not drawn in the figure, is also considered as a rigid body, which is standing on the ground surface and does not move. The distance from O to k is denoted by l_2 . Note that ΔT_k stands for the correction actuator torque, applied at joint k . Assuming that the additional torque ΔT_k will cause a change $\Delta\ddot{\beta}$ in acceleration of the upper part, while velocities will not change due to the action of ΔT_k , i.e., $\Delta\dot{\beta} \approx 0$. Next, the following equations are derived:

$$\Delta T_k = I_k \Delta\ddot{\beta} \quad (4.21)$$

$$F_z \Delta x = \Delta T_k + ml_1 l_2 (\cos \beta \cos \alpha + \sin \beta \sin \alpha) \Delta\ddot{\beta} \quad (4.22)$$

By virtue of Eq. (4.21), we have

$$\Delta\ddot{\beta} = \frac{\Delta T_k}{I_k} \quad (4.23)$$

Substituting Eq. (4.23) into Eq. (4.22), we obtain

$$\Delta T_k = \frac{F_z \Delta x}{1 + \frac{ml_1 l_2 (\cos \beta \cos \alpha + \sin \beta \sin \alpha)}{I_k}} \quad (4.24)$$

Taking into account that ΔT_k is derived by introducing certain simplifications, an additional feedback gain K_{zmp} is introduced into Eq. (4.24); thus Eq. (4.24) becomes

$$\Delta T_k = K_{zmp} \frac{F_z \Delta x}{1 + \frac{ml_1 l_2 (\cos \beta \cos \alpha + \sin \beta \sin \alpha)}{I_k}} \quad (4.25)$$

where K_{zmp} can be decided by the feedback in the actual walking. Equation (4.25) shows how to drive the actual ZMP towards the desired ZMP by controlling the torque output of the trunk joint.

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

In some special cases (for example, if the required torque is bigger than what the actuator of the trunk joint can provide), the actual ZMP may move out of the support area, which will make the exoskeleton intend to rotate around the edge of the foot. The user can feel such an intention because of the force transformed from the exoskeleton, and the user will control the exoskeleton walk an extra step; then, the support area is changed, and the ZMP falls back in the safe area; thus the rotation is stopped and the system recovers stable.

4.6 Sensory System of the Footpad

The sensory system of the exoskeleton's foot is used to measure the force interactions between the human operator and the environment. This information will be sent to the PCI card for the construction of various control algorithms, which will be discussed in Chapter 7. Thus, sensor selection for the sensory system is a crucial part in making a reliable exoskeleton. A variety of sensors, including six-DoF force-torque sensors, load cells and strain gauges, etc. can be employed under the exoskeleton's foot to obtain the ZMP information. However, each of these choices has its pros and cons. An ideal sensor should provide the following characteristics:

1. Light weight and minimal additional thickness under the foot
2. Provide a cost-effective solution
3. Able to withstand a weight of up to 40 kg.
4. Provide a real-time response to the control system during the walking motion.

4.6.1 Introduction of Selected Sensor

With its paper-thin construction, flexibility and force measurement ability, Flexiforce sensors are employed in the development of ZMP acquisition task. In this section, the characteristics and the working principle of the selected sensor will be discussed.

Figure 4.8 shows that the Flexiforce sensor is an ultra-thin (2 mm), flexible printed circuit. It is 0.55" (14 mm) wide and 8" (203.2 mm) in length. The active sensing area is a 0.375" diameter (9.53 mm) at the end of the sensor. The sensors are constructed with

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

two layers of substrate, such as a polyester film. The silver circle on top of the pressure-sensitive ink defines the active sensing area. Silver extends from the sensing area to the connectors at the other end of the sensor, forming the conductive leads. The two outer pins of the connector are active and the center pin is inactive.

Figure 4.8: Example of Flexiforce sensor

There are many ways to integrate the sensor into an application. One way is to incorporate it into a force-to-voltage circuit. A means of calibration must then be established to convert the output into the appropriate engineering units depending on the setup.

4.6.2 Working Principle of the Flexiforce Sensor

Figure 4.9: Excitation circuit

The Flexiforce sensor is a single element sensor that acts as a resistor in an electrical circuit. The schematic excitation circuit is shown in Figure 4.9. The formula for calculating output voltage is given as $V_O = -V_T(R_2/R_1)$. A negative 5 V as V_T is recommended. The

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

recommended R_2 is about $20\text{ k}\Omega$. When no load is applied to the Flexiforce sensor, the resistance is approximately $20\text{ M}\Omega$, whereas the resistance is $\geq 20\text{ k}\Omega$ if a full load is applied to it.

4.6.3 Sensor Calibration

The information and specification provided by the manufacturer is insufficient to ensure the sensors work accurately. Therefore, a calibration experiment is conducted to investigate the variation of the voltage with the applied load. Thus, an equation is obtained for ease of ensuing programming control.

Figure 4.10: The physical setup of the calibration

The calibration is carried out by means of the testing device shown in Figure 4.10 called *Instron*TM¹. This machine is fully integrated with a software package called Merlin, which provides a wide range of testing such as compression, tension, cyclic testing, etc.

Two out of eight sensors in the exoskeleton were randomly picked and labeled as sensor 1 and sensor 2. Table 4.1 shows the voltage differences with respect to the applied forces.

It appears from Figure 4.11 that the relationship between the voltage differences and the applied forces for both sensors is not varying too much where the forces are below 400N. Therefore, a general equation based on the two sensors is generated in Figure 4.12. The difference in voltages and applied forces of Flexiforce sensor can be related by $V = 0.0123F + 0.0807$, where F is the applied force and V is the voltages difference.

¹Instron testing machine, Aug. 2003, <http://www.instron.com/products/universalmaterial/5500.asp>

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

Table 4.1: Data recorded for sensors 1 and 2

Sensor 1				Sensor 2			
Test 1		Test 2		Test 3		Test 4	
Load (N)	Voltage (V)	Load (N)	Voltage (V)	Load (N)	Voltage (V)	Load (N)	Voltage (V)
0	0	0	0	0	0	0	0
56.0	0.66	58.4	0.71	55.7	0.79	56.7	0.75
108.6	1.36	107.4	1.37	109.0	1.52	109.1	1.37
158.2	1.99	157.9	1.92	160.1	2.10	160.1	2.22
205.3	2.52	206.5	2.47	206.2	2.60	205.9	2.86
254.2	3.14	254.2	3.01	255.5	3.14	255.4	3.55
307.2	3.82	307.8	3.72	307.1	3.67	307.1	4.14
359.0	4.46	360.2	4.27	356.4	4.25	356.3	4.81
397.7	4.90	411.5	4.83	408.8	4.81	409.1	5.37

Figure 4.11: Calibration result for Sensor 1 and 2

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

Figure 4.12: Generalize equation from Sensor 1 and 2

4.6.4 Initial Experiments

As shown in Figure 4.13, a test fixture and a footpad with sensors have been fabricated and some initial experiments have been performed.

Figure 4.13: The test fixture and footpad

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

(a)

(b)

(c)

Figure 4.14: Initial experiments

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

Experiments of controlling the exoskeleton's leg using the position signals measured from the human leg have been performed. As shown in Figure 4.14(a), encoders are attached to the human joints to measure the human leg movements and the exoskeleton leg is controlled to perform similar movements. Figures 4.14(b) and (c) are two snaps of the gait. Figure 4.14(b) shows the exoskeleton leg in swinging phase, while Figure 4.14(c) shows its support phase.

Figure 4.15: Signals from sensory system when tested on the foot alone

In analyzing the result obtained, a main factor that has to be taken into consideration is filtering. The signal appears acceptable (insignificant disturbances) when the sensory system is used to test the foot alone. As can be seen from Figure 4.15, the signal waveform before filtering is still acceptable and the Fast Fourier Transform (FFT) graph does not have any sharp spike. However, when the sensory system interfaces with the servomotor, the signals become unstable. This phenomenon is supported by the display shown in Figure 4.16, in which the signals before filtering are full of noise and the FFT graph have sharp spikes at higher frequency. The reason is the interference of high frequency vibration induced by the servomotor. Therefore, low pass filters are introduced prior to data processing.

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

Figure 4.16: Signals from sensory system when tested with the servomotor

Figure 4.17: The recorded ZMP trajectory and ground reaction forces acting on the exoskeleton's foot

CHAPTER 4. CONTROL STRATEGY OF THE EXOSKELETON SYSTEM

The resulting ZMP trajectory of this cycle is shown in Figure 4.17. The display output consists of two single-gait cycles. Each complete cycle consists of a negative peak and a positive peak. The negative peak indicates *heel contact* and the positive peak indicates *heel off*. Red path represents the exoskeleton's ZMP trajectory and the black path represents the operator's ZMP trajectory during the experiment. From the recorded trajectories, it is clearly shown that the ZMP trajectory of the exoskeleton (red path) is able to follow closely with the path of the human operator (black path). Apart from the trajectories history, the reaction forces acting on the exoskeleton are also recorded. It is shown that the forces change with the leg locomotion.

4.7 Summary

This chapter introduces the proposed control strategy including locomotion control and ZMP control in detail. Dynamics of the exoskeleton system is explored. A special footpad is designed, which can measure the ZMP of the human as well as that of the exoskeleton in real time. Using the measured human ZMP as the reference, the exoskeleton's ZMP is controlled by trunk compensation so that the exoskeleton can walk stably. Compared to other systems, this human-machine interaction method is easier for the user to operate the exoskeleton and no major inconvenience is caused.

Chapter 5

Simulation

It is necessary to verify a control algorithm using a software simulator before it is put into practice, because an unsuccessful control method may cause great damage to the user and/or the exoskeleton. On the other hand, to develop a hardware exoskeleton with the desired physical capability, it is necessary to clarify the required specifications of components in advance. An exoskeleton simulator, consisting of component parameters, is a useful tool to solve this problem. In this chapter, a human and an exoskeleton simulators are established using ADAMS ¹ and MATLAB ², and walking simulations are performed to verify the control algorithms. Moreover, the moving range of each joint and the power needed by each actuator are analyzed.

5.1 Previous Work

Motion control of articulated figures in virtual environments has been a popular research area in computer animation for years; these techniques are roughly classified into five categories: kinematics, dynamics, hybrid, constrained optimization, and motion editing. A brief summary of these techniques is presented next, with emphasis on algorithms that have been applied to human walking, especially those used for motion control of limb trajectories.

¹Adams University, Jan. 2005, <http://university.adams.com/tutorials.htm>

²MATLAB, Jan. 2005, <http://www.mathworks.com/products/matlab/>

CHAPTER 5. SIMULATION

5.1.1 Kinematics Approaches

Kinematics approaches for simulating human locomotion have been described by several researchers such as Calvert, Thalmann, *et al.* [59, 60, 61, 62]. These approaches for articulated figure animation can be further divided into two categories: *forward kinematics* and *inverse kinematics*. Forward kinematics approaches provide motion control by specifying the joint angles over time. The motion of the end-effector is determined as the accumulation of all transformations from the chain root to the end-effector. Inverse kinematics for end-effector goal positioning is adopted from robotics. It computes the joint angles for each segment in the chain structure from the position and orientation of the end of the limb.

5.1.2 Dynamics Approaches

Dynamics approaches describe the motion by a set of forces and torques from which kinematic data are derived. Dynamic simulation and control algorithms have been used to generate motions of articulated figures for years. There is a large number of robotics research concerning the control of bipedal locomotion, as well as biomechanics for simulating human walking motions [63, 64, 65, 66, 67, 68, 69].

5.1.3 Hybrid (Kinematics and Dynamics) Approaches

Beyond kinematics methods, some hybrid locomotion techniques have been proposed to generate walking motions by adding physical properties. The task is to find effective combinations that generate realistic motions while providing the animator reasonable and intuitive control over the motion. In general, simplified dynamic models are applied to simulate some parts of an articulated figure, such as the swing leg, support leg, or the body as a whole. Kinematics, on the other hand, gives animator the flexibility to control the desired motions. Girard, *et al.* [70, 71, 72] have implemented this technique in articulated figure animation.

5.1.4 Constrained Optimization

Constrained optimization approaches generate animation through an optimization of the objective subject to the constraints specified by the animator. Modeling the coordinated

CHAPTER 5. SIMULATION

articulated figure motion is fundamentally a problem of control, due to the nonlinear relationship between joint motions and limb movement and the need to satisfy constraints on a movements trajectory, speed, and energy expenditure. Furthermore, empirical studies of coordinated animal motion suggest that limb trajectories and body movement seem to be formulated in terms of performance optimization, such as minimization of jerk about the end of the limb [73].

Constrained optimization techniques have shown to be able to automatically generate expressive and natural limb motions that satisfy several of the basic principles of animation. Enhanced space-time techniques [74, 75] are especially suitable for complex motions. For example, locomotion on rough terrain could be broken into multiple space-time windows to satisfy the constraints and animation goals. However, the motions generated are highly depended on the animators ability to program the objective functions that meet the goals of a desired animation. Unfortunately, finding proper objective functions and formulating them for certain motions has appeared to be a difficult task for the animators.

5.1.5 Motion Capturing and Editing

An alternative way to obtain movements of articulated figures is to capture the motions from live subjects. Postures or motion sequences can be obtained with motion capture, and they can later be reused/modified and combined with editing tools. The complexity of the human figure and the limitations of current motion control systems, coupled with the increasing popularity and maturity (especially the hardware) of motion capturing, have made motion editing techniques [76, 77, 78, 79] become the recent trend of human animation.

5.2 Simulation Environment

To predict the physical biped (human/exoskeleton) motion, it is necessary to accurately formulate and solve the kinematic and dynamic equations of the mechanisms. However, the biped is a system with multiple bodies and many degrees of freedom. The analytical complexity of nonlinear algebraic equations of kinematics and nonlinear differential equations of dynamics makes it practically impossible to obtain closed-form solutions. On the other hand, there are some commercial dynamic software packages such as ADAMS, 3D

CHAPTER 5. SIMULATION

Working Model, DADS, etc., whose usefulness has been illustrated in many areas. These software packages can automatically formulate the equations of kinematics and dynamics, solve the nonlinear equations, and provide computer graphics output of the simulation results [80].

After comparison with other commercially available simulation software, ADAMS was selected due to its high numerical accuracy. Another benefit of ADAMS is that it incorporates easily w control software packages such as MATLAB, and MATLAB accepts user defined subfunctions. It is therefore possible to model detailed functions, define special constraints, and develop original control algorithms.

5.3 Human Model Representation

Figure 5.1: The human model

This section describes the kinematic structure of the human model. Figure 5.1 shows the human model established in ADAMS, which consists of legs, a waist, a trunk, and arms and a head. In reality, soldiers' hands are usually not free, and they should hold guns or something else. To simulate that situation, the human model holds a stick in its hands.

CHAPTER 5. SIMULATION

Model parameters such as link shape, size, mass and inertia can be arbitrarily designated. This model adopts measured human data and parameters are set according to an adult with a 65 kg weight and 1.7 m height. The lengths of the leg's parts, which are crucial parameters of walking, are listed in Table 5.1. There are three degrees of freedom in the hip joint, one in the ankle joint and one in the knee joint. There is also a DoF between the trunk and the waist.

Table 5.1: Parameters of the human model's lower limbs

Height of Ankle	Length of Calf	Length of Thigh	Length of Foot	Width of Foot
80 mm	340 mm	440 mm	200 mm	100 mm

5.4 Exoskeleton Model Representation

As shown in Figure 5.2, based on the human model, an exoskeleton model is established, and a payload is carried by the exoskeleton. The links of the exoskeleton are parallel to those of the human, and the height of each joint (ankle, knee and hip) is equal to the corresponding one of the human. The exoskeleton model is set to have the same size and mass as those of the actual exoskeleton that we designed and proposed to build up. The mass of each part of the exoskeleton is listed in Table 5.2. The details of the structure of the exoskeleton will be presented in next Chapter.

Table 5.2: Mass of the exoskeleton

Foot and Ankle Joint	Calf	Thigh and Knee Joint	Waist Frame and Two Hip Joints	Back Frame and Linear Actuators
3 kg	1.68 kg	4.77 kg	9.54 kg	6.12 kg
Grand Total	$(3+1.68+4.77)*2+9.54+6.12= 34.56$ kg			

CHAPTER 5. SIMULATION

Figure 5.2: The exoskeleton model

5.5 Walking Simulation

To run the walking simulation, three problems should be solved, namely, contact model, locomotion control and trunk compensation control.

5.5.1 Contact Model

The reaction force between the feet and the ground strongly affects the behavior of the exoskeleton and human model. For example, if the impact force arising when the feet hit the ground is too large, the burden on joints of supporting the exoskeleton becomes large, and the exoskeleton will easily vibrate and become unstable.

ADAMS adopts a spring-damper element to model the normal force between two contact objects. The IMPACT function activates when the distance between the first object and the second object falls below a nominal free length x_1 , that is, when two parts collide. As long as the distance between the first object and second object is greater than x_1 , the force is zero. The force has two components, a spring or stiffness component and a damping or viscous component. The stiffness component is proportional to the stiffness k , and is a function of the penetration of the first object within the free length distance from the second object. The stiffness component opposes the penetration. The damping component of the force is a function of the speed of penetration. The damping opposes the direction of

CHAPTER 5. SIMULATION

relative motion. To prevent a discontinuity in the damping force at contact, the damping coefficient is, by definition, a cubic step function of the penetration. Thus, at zero penetration, the damping coefficient is always zero. The damping coefficient achieves a maximum, c_{max} , at a user-defined penetration, d . The equation defining IMPACT is:

$$IMPACT = \begin{cases} \max(0, k(x_1 - x)^e - STEP(x, x_1 - d, c_{max}, x_1, 0) * \dot{x}) & : x < x_1 \\ 0 & : x > x_1 \end{cases}$$

To absorb the shock when the exoskeleton's feet contact the ground, we propose to install elastomeric material on the feet of the exoskeleton. This situation can be simulated to investigate the effect of this measure on the ground reaction force.

(a) Stiffness: 10^4 N/mm

(b) Stiffness: 10^3 N/mm

Figure 5.3: Modifying the stiffness

As shown in Figure 5.3, the stiffness of the exoskeleton's feet can be modified. Figure 5.4 shows the two ground reaction force curves of the exoskeleton's right foot corresponding to the two walking with different footpad stiffness. It can be seen that with the stiffness decreasing, not only the impact force is smaller, the force curve is also smoother. The simulation demonstrates that the elastomeric material can low-pass filter shock impulse and reduce the impact, thereby allowing the structure interacting with the environment at a high speed.

CHAPTER 5. SIMULATION

(a) Force at right foot when stiffness is 10^4 N/mm

(b) Force at right foot when stiffness is 10^3 N/mm

(c) Compare of the two force curves

Figure 5.4: Effect of foot's stiffness on the contact force

CHAPTER 5. SIMULATION

5.5.2 Locomotion Control

The human motion capture data (HMCD) is used to generate human-like walking motions for the human and exoskeleton models. As mentioned above, the human model's parameters are set according to an adult. To obtain the HMCD, encoders are attached to this adult at the joints of his lower limbs and the walking motion is captured.

The human limbs are soft due to the muscles and tendons, while the model's limbs are firm. Besides, the human has much more DoF than the model; hence, the human motion data cannot be used directly to drive the human model. The recorded data are processed and edited based on the inverse kinematics of the human model and interpolated using Akima spline before being input into ADAMS. Figure 5.5 shows the splines that are input to the human model joints, where the X axis represents time and the Y axis represents an angle. ADAMS will drive the human model's joints according to those trajectories.

5.5.3 Trunk Compensation

As mentioned in Chapter 4, control of the exoskeleton is divided into leg trajectory control and ZMP control. In this simulation, the exoskeleton model's leg trajectories are controlled by human model's leg trajectories. To keep the exoskeleton walk stably, the trunk should be controlled according to the ZMP of the exoskeleton. The trunk compensation function is implemented in MATLAB.

Figure 5.6 shows the block diagram in the MATLAB toolbox Simulink. It shows the flow of the signals during the simulations. At each time interval, the angular values, velocities and accelerations of the models joints sensed by the virtual sensors of ADAMS are sent to MATLAB, and those data are used by a trunk compensation module to calculate the torque needed at the trunk joint and such a torque will be applied to the virtual actuator of ADAMS, so that the exoskeleton model can walk stably in ADAMS. In other words, ADAMS provides the “hardware” of the exoskeleton model—virtual sensors, actuators and reaction forces between the model and the environment; while MATLAB provides the control software of the model.

Figure 5.7 shows the torque applied to the exoskeletons trunk joint for the changing masses of the load (5 kg and 20 kg). Figure 5.7 has special reference to selection of the suitable actuator for the trunk joint.

CHAPTER 5. SIMULATION

(a) Left ankle

(b) Right ankle

(c) Left knee

(d) Right knee

(e) Left hip

(f) Right hip

Figure 5.5: Akima splines

CHAPTER 5. SIMULATION

Figure 5.6: The block diagram in Simulink

CHAPTER 5. SIMULATION

mass of load = 5kg

mass of load = 20kg

Figure 5.7: Torque at the trunk joint

CHAPTER 5. SIMULATION

5.5.4 Simulation Results

Figure 5.8: Snapshots of the models while walking

Walking simulations have been performed. Figure 5.8 shows the snapshots of the models while walking in a simulation.

The successful simulations not only verified the proposed control algorithm, but also provided valuable information for the ensuing construction of the exoskeleton prototype. The information includes the moving range of each joint, and the torque and the power needed by each joint. Without loss of generality, we will only discuss the joints at the right leg of the exoskeleton. The payload is 20 kg in the following simulation.

CHAPTER 5. SIMULATION

(a) Heel off

(b) Toe off

(c) Mid swing

(d) Heel contact

(e) Foot flat

(f) Mid stance

(g) Heel off

Figure 5.9: Positions of the legs over a single gait cycle

CHAPTER 5. SIMULATION

Figure 5.9 shows the positions of the legs and corresponding joint angles during a single gait cycle from right heel off to right heel off, where the right exoskeleton's leg is denoted by red color. Below is the analysis of each joint.

Ankle

Although Figure 5.9 shows a small range of motion while walking (approximately -0.2 rad to 0.45 rad, or -11° to 26°), greater ranges of motion are required for other movements. An average person can flex their ankle anywhere from $-20^\circ \sim -30^\circ$ to $30^\circ \sim 50^\circ$ [13]. The LEE ankle was chosen to have a range of motion from -20° to 35° , which is greater than walking requires, but smaller than human anatomy allows in order not to hurt the user. Similarly, the motion range of knee joint and hip joint is designed also based on this criterion: larger than normal walking requires but smaller than human anatomy allows.

Figure 5.10: Ankle flexion/extension torque

Figure 5.10 shows the data of the ankle flexion/extension torque. The ankle torque is almost zero when the right leg is in swing phase. The torque is largest when the right foot is in its heel off phase.

The instantaneous ankle mechanical power is calculated by multiplying the joint angular velocity and the instantaneous joint torque, as shown in Figure 5.11.

CHAPTER 5. SIMULATION

Figure 5.11: Ankle instantaneous mechanical power

Knee

The knee buckles momentarily in early stance to absorb the impact of heel strike, then undergoes a large flexion during swing. This knee flexion decreases the effective leg length, allowing the foot to clear the ground when swinging forward. Although the walking knee flexion is up to approximately 0.6 rad or 34° , the human has significantly more flexibility—up to 160° when kneeling [13]. The LEE knee motion range was set to from 0° to 110° .

Figure 5.12: Knee flexion/extension torque

Figure 5.12 shows the data of the knee flexion/extension torque. The highest peak torque occurs in the early stance phase (from heel contact to foot flat). Figure 5.13 shows

CHAPTER 5. SIMULATION

the instantaneous mechanical power required by the knee joint.

Figure 5.13: Knee instantaneous mechanical power

Hip

The thigh moves in a sinusoidal pattern with the thigh flexed upward at heel contact to allow foot contact the ground in front of the person. This is followed by an extension of the hip through most of the stance phase and a flexion through swing. Figure 5.9 shows a small range of motion while walking (approximately -0.125 rad to 0.325 rad, or -7.2° to 18.6°). An average person can flexion reaches 90° with the knee extended. With the knee flexed, flexion can reach up to 120° or even beyond. When the knee is in extension, extension of the hip reaches 20° , when the knee is flexed, extension of the hip reaches 10° . The LEE hip angle is designed to 10° extension and 80° flexion. Figure 5.14 shows the data of the hip flexion/extension torque and Figure 5.15 shows the instantaneous mechanical power required by the hip joint.

CHAPTER 5. SIMULATION

Figure 5.14: Hip flexion/extension torque

Figure 5.15: Hip instantaneous mechanical power

5.6 Summary

In this chapter, the proposed control algorithms, including locomotion control and trunk compensation, are verified via walking simulations. Furthermore, the simulation results provide valuable information for the ensuing construction of the exoskeleton prototype. The moving range of each joint, and the torque and power needed by each joint are analyzed. The developed simulator, which consists of adjustable component parameters, is a useful tool for the further development of the exoskeleton or other biped robots.

Chapter 6

Design and Construction of Prototypes

This chapter introduces the design and fabrication of the lower extremity exoskeleton system, including the inner lower exoskeleton and the outer lower exoskeleton.

6.1 Inner Lower Exoskeleton

The inner exoskeleton is used to attach the encoders to the human limbs. It is located between the human limbs and the outer lower exoskeleton and its weight is carried by the user. Hence, it must be as light and compact as possible.

6.1.1 Conceptual Design 1

Figure 6.1: Conceptual design 1

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

The design of the rotary encoder bracket system was initially focused on the development for the knee linkage.

In the initial design, we aimed to hold the encoder in place on the linkages via the use of the standard issued steel clips. Attempt to reduce the thickness of the bracket was done via omission of the standard issued plastic coupling. Upon close observation, one would realize that the linkage on the right has its weight purely supported by the rotating spindle of the rotary encoder. This spindle is fragile by nature. Hence such a design was dismissed as being unfeasible.

6.1.2 Conceptual Design 2

For the second attempt, we diverted support of the lower linkage from the steel spindle to the upper limb via the use of a ball bearing. The bearing is secured to the lower linkage with the use of a C-clip while the upper linkage is secured via the use of the flange of the bearing.

The use of a coupling is not required in this design. The encoder is held in place via the use of an acrylic disc. The disc is then secured to the upper linkage via the use of four standard 3 mm bolts.

Figure 6.2: Conceptual design 2

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

6.1.3 Conceptual Design 3

Figure 6.3: The encoder bracket

Figure 6.4: The assembled encoder housings

This is the final design we selected. A minimum protrusion is achieved by designing the brackets to allow the tip of the encoder spindle to actually touch the surface of the user. This feat was achieved by the use of a larger but extra thin bearing which allows the rotary encoder to reside within the bearing itself. The encoder bracket is shown in Figure 6.3.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

In this design, linkages are separated from the encoder housing. This modular concept of the housing allows the same housing unit to be repeated for the hip unit, the knee unit and the ankle unit. This modular design also cuts down on both design and manufacturing time. The production cost is also reduced. The encoder housings with linkages are shown in Figure 6.4.

6.1.4 Inner Exoskeleton Harness Design

To position the rotary encoder brackets firmly on the respective bodily positions on the lower extremity of the human body, a need arise to design a harness system which is versatile, reliable, light and durable. Nylon being a material that economical, durable and easily worked with was selected as the choice material for the harness.

For the harness used in the knee brackets, Velcro was chosen due to its ability to conform readily to the shape of the user's thigh. Also, with the Velcro, we can easily compensate for slight variations in the dimensions of the various user thigh heights without having to adjust the buckle position.

Figure 6.5: The user wearing the inner exoskeleton

Figure 6.5 illustrates the feasibility of the inner exoskeleton when mounted on the user's

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

lower extremity. It can be shown that the inner exoskeleton does not constrain any motions or postures performed by the user.

6.2 Outer Lower Exoskeleton

The OLE provides the mechanical hardware means for the exoskeleton system. Therefore, it must be made of structures and mechanisms strong enough to carry a load and travel by itself. The OLE is a powered lower exoskeleton, which means that it is to be worn by the user from the user's waist down to the user's feet. The OLE has to be able to carry its own weight as well as a payload and should not hinder the walking movements of the user.

6.2.1 Requirements and Constraints

The requirements and constraints regarding the design of the OLE are briefly listed below:

Low cost of fabrication and assembly: Costs of building the OLE prototype must be low and economical.

Light weight: The overall weight of the OLE system must be as lightweight as possible. This is to ensure that the power needed from the actuators to power the OLE limbs will be kept minimal. As such, power consumption will be reduced.

High safety level: The OLE must be able to operate safely and not cause any injuries or pose any hazards to the user and the environment.

Easy to build: The OLE must be designed for easy machining and assembly.

Easy to wear: The user should be able to wear and subsequently disengage from the OLE whenever needed in an easy, transparent and quick method.

High aesthetic satisfaction: The OLE must look sleek, compact and at the same time, must "feel" sturdy and reliable. Also, it is preferred that the OLE will resemble the actual human lower limbs.

Low noise level: The OLE should not produce excessive noise while operating.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

6.2.2 Various Conceptual Design

Based on the criteria listed above, the various solutions to the sub-functions are discussed and compared, the most suitable solution is then selected.

(a) Types of Attachments

The OLE is to be attached to the user at waist point and at both feet. Importantly, the OLE has to be attached securely to the user such that the user's movements are not hindered. Features like quick-release attachments are also a plus. To attach OLE to the user at the waist, a few solutions are generated.

Figure 6.6 shows four solutions to attach the user at the waist.

Figure 6.6: Solutions to attach user at waist

Solution I: Waist Belt

Using a waist belt, the user can be attached to the OLE by fastening the belt at the waist using a simple buckle. This is shown in Figure 6.6 (a).

Advantages:

- Simple
- Cheap to make
- Comfortable

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

- Adjustable size
- Quick-release easily obtained
- No hindrance to user's movements

Disadvantages:

- Belt may slip after sometime
- Concentrated area of contact might cause abrasions or discomfort after prolonged periods

Solution II: Waist-Thigh Harness

In Figure 6.6 (b), the OLE can be attached to the user via a safety waist-thigh harness. These harnesses are often utilized in climbing sports and outdoor activities.

Advantages:

- Secure and firm, no slips
- No hindrance to user's movements as the belt is strapped firmly around the waist and thighs separately
- Easily acquirable

Disadvantages:

- Inconvenient to wear or strip, requires a longer time to wear or strip
- No quick-release features
- Uncomfortable to wear after some time, mainly due to abrasion and heat
- Danger of harness being caught by other parts of the OLE system

Solution III: Combat Vest

Figure 6.6 (c) shows a combat vest worn by soldiers during outfield activities. The combat vest can also carry other items like ammunition magazines, grenades, and other combat accessories.

Advantages:

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

- Secure and firm, no slips
- Able to carry many items
- Quick-release possible
- Large area of contact with the user's body, hence less chances of abrasion

Disadvantages:

- Difficult to integrate onto the OLE
- May get entangled with other parts of the OLE, hence a cause of hazard

Solution IV: Bullet-Proof Vest

Figure 6.6 (d) shows a bulletproof vest. The OLE can be attached to the user via the vest and also acts as protection to the user. This solution is especially relevant as the expected users of the OLE are supposed to be soldiers.

Advantages:

- Secure and firm, no slips
- Additional protection to the user
- Large area of contact with the user's body, hence less chances of abrasion

Disadvantages:

- Difficult to integrate onto the OLE
- Expensive
- No quick-release features

(b) Means of Limbs Adjustment

Due to human factors, the design of the OLE limbs must fit the sizes of different users. Various methods of easily adjusting the lengths are discussed below.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.7: Solutions to adjust length of OLE limbs to user size

Solution I: Clutches Adjustment Method

Utilizing the concept already used in clutches, Figure 6.7 (a) shows a solution to adjust the length of the OLE limbs. The method consists of two hollow tubes, one inserted within the other. On the outside tube, holes of equal intervals are drilled through. A circular spring ledge is attached on the inside tube. The OLE limbs can be customized by sliding the tubes to the required length. The two tubes can be fastened by aligning the holes at the outside tubes to the spring ledge on the inside tube.

Advantages:

- High reliability
- Easy to build
- Less number of parts

Disadvantages:

- Only adjustable to fixed intervals, thus insufficient in accuracy
- Can be difficult to align the spring ledge to the holes
- Often difficult to dislodge the spring ledge from the hole position, once it fits into the hole

Solution II: Screwing Method

In Figure 6.7 (b), the method makes use of a rod, a base and a nut. The rod is screw threaded and there is a tapped hole at the top of the base. The tapped hole has corresponding diameter to threaded rod. The length of the limb is adjusted by screwing the rod into the

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

base to attain the specific length. The rod is subsequently fixed by tightening the nut along the screw threads against the base.

Advantages:

- Accurate adjusting of length
- Easy to operate
- Simple mechanism
- Makes use of mechanical advantages

Disadvantages:

- Screw threads can be easily damaged
- May need the use of bearings to orientate the planar positions of the limbs
- More number of parts
- Difficult to make screw threads on thick rods
- Difficult to tap holes of large diameter
- Detailed and complicated calculations needed to determine the pitch needed for the threads

Solution III: Bicycle Clamp Method

Using the existing bicycle clamp method shown in Figure 6.7 (c), the OLE limbs can be adjusted to the user's height. The outer diameter of the inner tube must be slide-fit to the inner diameter of the outer tube when the bicycle clamp is released. When the bicycle clamp ledge is tightened, the outer tube should clamp the Inner tube tightly such that the two tubes will not slide under huge vertical forces.

Advantages:

- Reliable and strong as existing method can even withstand the weight of the whole human being when the bicycle clamp is ledged tight, as in the case of the bicycle seats
- Easily available, the bicycle clamp is a readily available standard part

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

- Easy to build (the two tubes)
- Cheap
- Able to adjust a full range of lengths, thus accurate match to the user's anthropometrics
- Need not worry about the orientation of the limbs as no turning actions are required of the OLE limbs when doing the length adjustments

Disadvantages:

- Tolerance may be difficult to enforce for the critical inner and outer diameters
- Prone to wear and tear after frequent adjustments as the sliding motion between the two tubes often causes excessive friction, and thus results in a gradual loss of materials from the tubes

(c) Methods of ZMP Adjustment

Actuators have to be added to adjust the payload so that the overall system of the OLE remains in equilibrium. To achieve this critical function, a few ways of mounting actuators to adjust the payload position about the lateral axis only will be discussed below.

Figure 6.8: Solutions to adjust payload to correct ZMP

Solution I: Linear Actuator On Top

The linear actuator shown in Figure 6.8 (a) is mounted on top of the waist platform. Because of the user's body position, the linear actuator has to be placed by the side of the

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

waist. The actuator adjusts the positions of the payload about the lateral axis using linear strokes.

Advantages:

- Linear actuators are sensitive enough and are able to provide small strokes
- No transmission is needed
- Simple and direct force application on the payload
- Easy to build

Disadvantages:

- Expensive
- Might hinder the arm movements of the user

Solution II: Linear Actuator Below

For Solution II, the linear actuator is placed below the waist platform and adjusts the payload from below. This is shown in Figure 6.8 (b). For this arrangement, the linear actuators can act at the center of the payload unlike the linear actuator in Solution I.

Advantages:

- Only one actuator needed
- Easy to build
- No transmission is needed
- Simple and direct force application on the payload
- Linear actuators are sensitive enough and are able to provide small strokes

Disadvantages:

- Bulky design
- Actuator on high bending stresses
- Might hinder the user's lower limb movements

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Solution III: Rotary Motor

In Figure 6.8 (c), the payload is adjusted by rotary motors. The rotary motor rotates the payload about the lateral axis as shown, and is thus able to adjust the payload positions accordingly.

Advantages:

- Compact
- Will not hinder the user's movements

Disadvantages:

- High torque needed
- Requires transmission

(d) OLE Actuators and Transmissions

A few actuators are chosen and compared. The comparisons between electromagnetic, hydraulic and pneumatic actuators were already drawn in the previous chapter. After studying each of these conventional actuators, we decided to streamline the choice of actuators to only electromagnetic actuators. This is mainly because electromagnetic actuators are easily available at competitive prices. Also, electromagnetic actuators can satisfy the need for position control in the OLE system. Wireless options in power supplies for electromagnetic actuators are also obtainable in the form of batteries. Ultimately, we feel that electromagnetic actuators are the simplest and most applicable option for the OLE.

The following solutions describe the various possible types of electromagnetic actuators and their respective transmissions. These combinations of actuators and transmissions are selected mainly to satisfy large torque requirements at the OLE joints and the preferred low power to weight ratios for the actuators.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.9: Solutions for OLE actuators and transmissions

Solution I: DC Servomotors with Gearboxes

DC servomotors remain one of the most easily available actuators in the market. With gearboxes, a higher torque can be obtained. The servomotor can be easily controlled in reference to position or torque and, with a gearbox, presents the most direct and simple method of actuation. Figure 6.9 (a) shows a configuration of a DC servomotor connected to a gearbox transmission.

Advantages:

- Relatively cheap compared to other forms of actuators
- Easy to control
- Easily available

Disadvantages:

- Bulky and heavy in size for servomotors with higher torque
- Loss in efficiency through the transmission of the gearbox
- More suitable for high speed, low torque applications, while the OLE joints need high torque at low speeds
- Low power to weight ratio

Solution II: Linear Actuators

Figure 6.9 (b) shows the previous work in building the OLE limb. Linear actuators were used to actuate the hip and knee joints as shown in the Figure.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Advantages:

- No additional transmissions needed
- Suitable for low speed applications

Disadvantages:

- Bulky, takes up a lot of space in the sagittal plane of the OLE limbs as the linear actuators form triangles with the other two links to actuate the joint
- Low power to weight ratio

Solution III: Harmonic Drive Actuators

Harmonic drives are powerful actuators using state-of-art harmonic gear technologies. Figure 6.9 (c) shows the FHA Series Hollow Shaft Servo Systems. This FHA Series integrates a servomotor with a harmonic gear drive as one and thus provides a very compact and powerful solution for actuating the OLE joints. Its flat and compact frame helps to make the OLE joints less bulky. More information regarding harmonic drives can be found in the website ¹.

Advantages:

- High power to weight ratio
- Compact
- No additional transmissions needed
- Capable of high torques and low speeds, especially suitable for the OLE joints
- Easy to control
- Aesthetically appealing

Disadvantages:

- More expensive than most servomotors

¹Harmonic Drive Systems, Dec. 2004, <http://www.hdsystemsinc.com/principles/>

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Concept Evaluation

Concept evaluation is done with respect to the evaluation criteria previously set up. After evaluating the advantages and disadvantages of each possible solution, this final conceptual design makes use of: the waist belt to attach user at waist; the bicycle clamp method to adjust the length of the OLE limbs to user size; the linear actuator on top to adjust payload to correct ZMP; and lastly, harmonic drives for the OLE actuators and transmissions, as shown in Figure 6.10. The final conceptual design is shown in Figure 6.11.

Figure 6.10: Morphological chart for the OLE

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.11: Final conceptual design

6.2.3 Embodiment Design

The embodiment design can be mainly divided into a few different modules. These modules include the waist assembly, hip joints, upper limbs, knee joints, lower limbs, ankle joints and lastly, the OLE feet. The following will introduce the main features of them one by one.

The ***waist assembly*** consists of the back frame, waist belt, linear actuators, as well as a waist platform for attachments.

The ***hip joints*** describe the actuation of the joint using the harmonic drives. In addition, a Rosta assembly is added at each hip joint for passive movements in the abduction and adduction directions.

The ***upper limb*** will be the linkage between the hip joint and the knee joint. Bicycle clamps will be included for length adjustments.

The ***knee joints***, like the hip joints, relate mainly to force transmission. However, no Rosta assemblies will be added.

The ***lower limb*** acts as a linkage between the knee and ankle joints. Similar to the upper limb, bicycle clamps are included for length adjustments.

The ***ankle joints*** actuate the ankles and will include force transmission using harmonic drives as well as the use of the Rosta assemblies for passive movements in the supination

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

and pronation directions.

Lastly, the *OLE feet* include force sensors to measure the ZMP of the exoskeleton and the pilot.

In all, there will be one waist assembly module, two hip joint modules, two upper limb modules, two knee joint modules, two lower limb modules, two ankle joint modules, and likewise, two OLE foot modules. Embodiment design on each different module will be discussed next.

Before the embodiment design, modifications are made to the final conceptual design described in the previous section mainly due to change in the actuator used. Because of budgeting problems, the original intention of using the FHA series harmonic drive actuators had to be replaced for the building of the OLE prototype. To build the OLE prototype, existing motors on the previous NTU robotics project, the Haro, are used to actuate the OLE joints. The Haro had actuator servomotors connected to harmonic drive gears. Because they are separate components, this configuration of actuator and transmission is not as compact as the FHA series of harmonic drive actuators mentioned earlier. However, the Haro actuators still present a viable solution for actuating the OLE joints. After dismantling the joints of the Haro, the original assembled modules of the actuators and harmonic gears are used for easy assembly with the respective OLE joints. To incorporate the Haro joints to the OLE joints, certain modifications to the OLE joints, especially to the mounting brackets, have to be made.

(a) Waist Assembly

Figure 6.12 shows different views of the waist belt. To attach the waist belt to the waist platform, a sliding attachment is designed. This attachment allows the waist belt to slide up and down for a small range and thus, giving the user some free play when wearing the waist belt. This is especially important as some vibrations and movements are expected from the waist belt as the user moves around. The embodiment design of this sliding attachment can be seen in Figure 6.13.

In Figure 6.13, the sliding plate can slide along the support shafts for a small range. A bracket, holding the support shafts, is also used to attach the whole sliding attachment to the waist platform. The waist belt will be attached to the sliding plate. All these are described in Figure 6.13. For wearing of the waist belt, a standard plastic buckle is used to

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

fasten the belt to the user's waist.

Figure 6.12: Waist belt

Figure 6.13: Sliding attachment

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.14: Waist assembly

Figure 6.14 shows the overall embodiment design of the waist assembly. It shows how the back frame, linear actuators, waist belt, and waist platform are assembled to form the waist assembly module. A few views are presented in Figure 6.14 to help visualize the embodiment design.

(b) Hip Joints

For the hip joint, one of the actuator and transmission modules from the Haro was chosen. The Rosta assembly is incorporated, by fastening it to the output link of the Haro actuator module, with a mounting plate and standard Rosta brackets. Basically, the rubber suspension in the Rosta assembly allows for slight passive movements of the hip joint in the abduction and adduction directions. Figure 6.15 describes the hip joint.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.15: Hip joint

Figure 6.16: Upper limb

(c) Upper Limbs

Other than providing support between the hip and the knee joint, the upper limb has adjustable length to suit different users. Figure 6.16 shows the upper limb. The top link is to be inserted into the base link through the bicycle clamp. When the bicycle clamp is tightened, the top link will be camped tightly within the base link. When the bicycle clamp is released, the top link can slide in and out the base link, varying the length of the upper limb. Subsequently, the top link will be connected to the hip joint and the base link will be connected to the knee joint.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

(d) Knee Joints

Figure 6.17: Mounting bracket of the knee joint

Figure 6.18: Actuated Link of the knee joint

Figure 6.17 shows the design of the mounting bracket. Designs to limit the degree of rotation of the actuated link are also included in the mounting bracket. Such mechanical stopper design is to protect human legs. Of course, the mounting holes have to be re-designed to suit the mounting holes already existing on the Haro module. For the actuated

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

link, mounting holes were redesigned as well to suit the Haro module and a triangular protrusion was added to stop the actuated link from overturning. Design of this new actuated link is shown in Figure 6.18. Figure 6.19 describes the overall design of the knee joint.

Figure 6.19: Knee joint

(e) Lower Limbs

The lower limb design comprises a base link and a bicycle clamp. Like the upper limb, the bicycle clamp is slotted at the top of the base link for length adjustment. The only difference between the upper limb and the lower limb is the absence of a top link. This is because, unlike the upper limb, the actuated limb of the knee joint is directly inserted into the base link of the lower limb.

(f) Ankle Joints and Feet

The ankle joint comprises the Haro module and Rosta assembly, as Figure 6.20 shows. The footpad is designed as described in Chapter 3. The basic concept is to use three layers of plates to incorporate the force sensors. On the first plate of the OLE footpad, velcro straps were fastened for the wearing of the user's foot. There is a rubber sole to absorb impact. Figure 6.21 shows the OLE foot connected to the ankle joint.

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.20: Ankle joint

Figure 6.21: Ankle joint connected to the foot

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

The fully OLE prototype looks like what is shown in Figure 6.22.

Figure 6.22: The overall design

6.2.4 Assembly and Implementation to the OLE

For all the fabricated parts, the material used mostly is aluminum, except for the sliding attachment in the waist assembly. The sliding plate on this sliding attachment (see Figure 6.13) was to be made of acrylic. This is because the main function of this plate is only for the attachment of the waist belt and not for any structural load bearing.

The OLE prototype was assembled successfully by February 2005. This final assembly of the OLE prototype can be seen in the Figures 6.23 to 6.27. Figure 6.23 shows the overview of the whole OLE prototype. Figure 6.24 shows the hip joint. Figure 6.25 shows

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

the upper limb and its bicycle clamp. Figure 6.26 shows knee joint. Lastly, Figure 6.27 shows the ankle joint with the OLE foot.

Figure 6.23: Photos of the OLE prototype

Figure 6.24: Photos of the hip joint

Figure 6.25: Photos of the limb

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

Figure 6.26: Photos of the knee joint

Figure 6.27: Photos of the ankle joint with OLE feet

CHAPTER 6. DESIGN AND CONSTRUCTION OF PROTOTYPES

6.3 Summary

This chapter discusses in detail the design and construction of the prototype of the exoskeleton system. The function of each part is explained clearly and various solutions are discussed and compared. Then, the most suitable solutions are selected and the prototype is fabricated and assembled. The prototype consists of seven positive DoF and four passive DoF. It is modular, low cost and adjustable for different users. Its weight is about 30 kg. The lengths of calf and thigh are adjustable from 280 mm to 430 mm, and 390 to 550 mm, respectively. The footpad of the exoskeleton is specially designed, which can measure the exoskeleton's ZMP as well as the user's ZMP in real time. The prototype constructed provides a testbed for the control algorithms.

Chapter 7

Walking Implementation of the Exoskeleton

The important function of the inner exoskeleton is to obtain necessary input data from the operator. The data will be analyzed, and transformed into corresponding command signals for the outer exoskeleton using certain mapping algorithms. An operation of a dynamic system is called a real-time operation if the combined reaction- and operation-time of a task is shorter than the maximum delay that is allowed. When the human operator performs one action, the OLE should react correspondingly as fast and precisely as possible, i.e., the OLE should be real-time controlled.

7.1 Real-Time Operating System Used in this Work

A Real Time Operating System or RTOS is an operating system that has been developed to serve for real-time applications. Some examples of RTOS are QNX, RMX, RT-11, LynxOS, MicroC/OS-II, etc.

In this work, xPC Target toolbox in MATLAB is employed. xPC Target provides a high-performance, host-target prototyping environment that enables developers to connect system simulation models (which can be created using Simulink and Stateflow toolbox from MATLAB) to physical systems and execute them in real time on PC-compatible hardware. xPC Target provides comprehensive software capabilities for rapid prototyping and hardware-in-the-loop simulation of control and signal processing systems. It also enables users to add I/O interface blocks to their models (system models and controller designs etc.), automatically generate code with Real-Time Workshop and Stateflow Coder, and

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

download the code to a second PC running the xPC Target real-time kernel. This capability to rapidly produce code directly from verified system results allows many different techniques to be tried on hardware in a more convenient way compared to other operating systems. Developers who use the other systems, most of the time, have to deal with low-level programming languages directly. These kinds of low-level programming languages are somewhat not very efficient in the sense that they take developers long time to deal with coding and debugging.

The system includes software such as Matlab, xPC Target etc., which run in the PCs, and I/O boards that receive commands from xPC Target and control the motors via amplifiers.

Figure 7.1: xPC Target system [81]

The xPC Target system runs on two PCs, host and target, as shown in Figure 7.1. The host PC runs Microsoft Windows 2000 operating system and the required software packages: MATLAB, Simulink, Real-Time Workshop (RTW), xPC Target, and C/C++ compiler. The Microsoft Visual C++ compiler is used in this work. MATLAB is the host software environment of Simulink, RTW, and xPC Target. The command-line interface can be used to control and interact with the xPC Target software environment and target application. Simulink is used to model dynamic physical systems (exoskeleton systems in our work) and controllers using block diagrams. RTW and C/C++ compiler convert Simulink blocks into C code and build a target application, which is then downloaded to the target PC and executed in real-time. The communication between the host PC and the target PC is accomplished through RS-232 since it is inexpensive and easy to install. The target PC is booted by a boot disk. Subsequently, the target PC runs the highly optimized xPC target kernel loaded from the boot disk. The kernel initiates the host-target communication,

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

activates the application loader, and waits for the target application to be downloaded from the host PC. The executable code generated from the host PC then runs in real-time on the target PC.

The overall architecture of the host-target hardware-in-the-loop exoskeleton control system established in this work is shown in Figures 7.2 and 7.3.

Figure 7.2: Control system architecture of the exoskeleton system

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.3: Developed control system set up

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

7.2 Control of the Developed Exoskeleton

Figure 7.4: Hierarchical Architecture of Control Biped Locomotion Robots [1]

In general, hierarchy is a basic principle on which large scale systems are controlled. The control structure most often applied on bipedal robots is of four levels [1], as shown in Figure 7.4:

- 1) Highest control level: recognizes the obstacles in the operating space and the conditions under which a task is being performed, and makes the decisions on how the task imposed is to be accomplished.
- 2) Strategic control level: divides the imposed operation into elementary movements.
- 3) Tactical control level: distributes an elementary motion to the motion of each DoF of the robot.
- 4) Executive control level: executes the imposed motion of each DoF.

All robots have the two lowest control levels. The tactical level generates the trajectories of each DoF, which perform the desired functional movement, while the executive level

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

executes these trajectories by means of appropriate actuators, incorporated in each DoF. The two upper control levels are generally recognized as intelligence level [1]. Robots with the two upper levels such as ASIMO [82], HRP-2 [83, 84], etc. can walk on unknown terrain. Robots walking in a known environment whose trajectories of each link can be pre-defined off-line do not need the two upper levels. Instead of artificial intelligence, the highest level of the exoskeleton's control structure is implemented by the human user. The user recognizes the obstacles using his/her natural sensors, decide where to go and how to go. The strategic control level of the exoskeleton's control structure is to transfer the human intention into the exoskeleton's controller and the controller divides the imposed operation into elementary movements. The two lowest control levels are similar to those of the robots. In other words, the exoskeleton can be seen as a robot integrated with the human intelligence.

In this section, the two lowest control levels are first tested to ensure that the joints of the exoskeleton can perform the desired trajectories in real time. The whole control strategy is then implemented and experiments of walking with the pilot are performed.

7.2.1 Joint Control

Figure 7.5: Block scheme of OLE with controller for testing

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.5 shows the block scheme of controlling OLE's joints in the MATLAB's Simulink. Two I/O boards are employed in this work. Note that for each I/O board, the initialization block should be included in the Simulink block diagram. These I/O board blocks actually contain the specific software code (drivers) to establish the interface between the Sensoray Model 626 I/O boards with the target computer. The I/O board, which is plugged into the PCI slot of the target PC, connects the amplifier and encoder through proper channels. Each board reads six encoders and controls the three motors of one leg. The upper one is to control the right leg and the lower one is to control the left leg. The channels 1 (hip), 2 (knee) and 3 (ankle) read the counts from the encoders of the inner exoskeleton, while channels 4 (hip), 5 (knee) and 6 (ankle) read the counts from the encoders of the OLE. Then the counts are converted into degrees and compared. The difference between the inner and outer exoskeleton's corresponding joint is converted to a command signal through the PID controller. Then, suitable voltage will be applied to the amplifiers through the channels (1 hip, 2 knee, 3 ankle) of the Sensory Analog Output block and motors will actuate the OLE to move following with the inner exoskeleton.

Figure 7.6 shows the response of OLE's left knee joint during one walking test, where the horizontal axis represents time (second), the vertical axis represents angular displacement, the blue curve represents the inner exoskeleton, the green curve represents the OLE, and the red curve represents the difference between the inner exoskeleton and the OLE. Figure 7.7 shows the difference in an enlarged scale. Similarly, Figures 7.8 and 7.9 show the response of OLE's hip joint and ankle joint, respectively. It can be seen from those figures that the OLE responds fast and the difference is small, i.e., the joints are real-time controlled.

Figure 7.6: Response of the left knee joint

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Angle difference (degree)

Figure 7.7: Angle difference between the inner exoskeleton and the OLE

Figure 7.8: Response of the left hip joint

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.9: Response of the left ankle joint

Figure 7.10: Exoskeleton trial with post-stroke patient

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.11: Exoskeleton trial with a normal user

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.10 shows the snapshots of a trial with a user whose left leg is very weak after strokes; and Figure 7.11 shows the snapshots of another trial with a normal user. Both users confirmed that the exoskeleton's leg responds well, as they could hardly feel any extra burden.

7.2.2 Measurement of the ZMP

As mentioned in Chapter 4, to make the exoskeleton walk stably, ZMP should be introduced into the control strategy. In this section, the experiment of measuring the human ZMP during walking is presented.

In this experiment, a human walks four steps, as shown in Figure 7.12. First, the human stands up, the width between his left foot and right foot is 300 mm. Then he walks forward, and his right foot moves first, from position R1 to position R2. Next, his left foot moves from position L1 to position L2. Thirdly, his right foot moves from position R2 to position R3. Lastly, his left foot moves from position L2 to position L3 and he recovers the standing posture. The step length is 300 mm.

There are four force sensors below each foot of the human. The reading of each force sensor below the left and right foot and their grand total are shown in Figure 7.13 and Figure 7.14, respectively. Figure 7.15 shows the timing of the ground reaction force below the two feet. At the first second, the human begins to walk. From the fifth second the human recovers the standing posture.

Using Eqs. (4.17) and (4.18), the ZMP of the human is calculated, as shown in Figure 7.16. Figure 7.15 and Figure 7.16 show that when the human begins to walk, the ground reaction force below his right foot decreases while the force below his left foot increases. The ZMP shifts from the middle of two feet to the left foot, i.e., the supporting foot area. During walking, the ZMP shifts between the two feet. Lastly, when the human recovers the standing posture, the ZMP stays between the two feet.

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.12: Steps of the walking experiment

Figure 7.13: Ground reaction force below the left foot

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.14: Ground reaction force below the right foot

Figure 7.15: Timing of the ground reaction force below the two feet

Figure 7.16: ZMP trajectory in the X-Y coordinate system

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

7.2.3 Exoskeleton's Walking Experiments

Figure 7.17: Signal flow diagram

Figure 7.18: Block scheme for walking

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

An online balancer is added into the control scheme, as shown in Figure 7.17. Figure 7.18 shows the block scheme in the MATLAB's Simulink. One more I/O board is employed, which reads the encoder of the trunk joint, signals from force sensors and controls the actuator of the trunk joint. The balancer module reads the current joint angles and measured human ZMP, then calculates the desired OLE's ZMP. If the measured OLE's ZMP differs from the desired one, a suitable command signal will be applied to the actuator of the trunk joint and trunk compensation will shift the actual ZMP toward the desired position, as described in Chapter 4. The ZMP will stay within the supporting area and the OLE will walk stably.

Figure 7.19: Human and exoskeleton's ZMP trajectories in the X-Y coordinate system

A walking experiment is performed to verify the balancer module. The walk procedure is similar to that in Figure 7.12. Figure 7.19 shows human and exoskeleton's ZMP trajectories in the X-Y coordinate system. This figure shows that the ZMP is adjusted during the single support phase and the ZMPs are always kept in the support area. Hence the exoskeleton can walk with the user stably.

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

Figure 7.20: X coordinate of the human and exoskeleton ZMPs with respect to time

Figure 7.21: Y coordinate of the human and exoskeleton ZMPs with respect to time

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

To see the relationship between the human ZMP and exoskeleton's ZMP more clearly, Figures 7.20 and 7.21 show the X and Y coordinates of the human and exoskeleton ZMPs with respect to time, respectively. These figures show that, in the X direction, the exoskeleton's ZMP is close to the human ZMP due to the trunk compensation, which ensures that the exoskeleton can walk stably following the human. In the Y direction, the distance between exoskeleton's ZMP and the human ZMP is bigger. This is because the exoskeleton's footpad is much wider than human footpad (see Figure 7.19); therefore, the allowable range of the exoskeleton's ZMP in Y direction is bigger.

More experiments are performed that include walking forward and backward with a payload of 20 kg. Figure 7.22 shows some snapshots of one of those experiments.

Figure 7.22: Snapshots of a walking test

CHAPTER 7. WALKING IMPLEMENTATION OF THE EXOSKELETON

7.3 Summary

In this chapter, the real-time control is implemented on the prototype constructed. xPC Target, together with other toolboxes from MATLAB are used to provide a real-time operating system and an integrated development environment. Experiments of joint control are firstly introduced to show that the joints of the exoskeleton can perform the desired trajectories in real time. The user confirmed that the exoskeleton's leg responds fast, as he could hardly feel any extra burden. After that, the control strategy introduced in the previous chapter is implemented. Experiments show that the exoskeleton, which carries a payload of 20 kg, can walk with the pilot at a speed of 0.3 m/s or one step per second. It can be seen from the recorded ZMP trajectories that the exoskeleton's ZMP is controlled to follow the pilot's ZMP closely.

The developed prototype together with its control system provides a testbed for study of the exoskeletons. Based on the works we have done, the development of a new exoskeleton, which is more compact and comfortable, is ongoing.

Chapter 8

Conclusions and Recommendations

8.1 Conclusions

This thesis includes the design, development and control of a wearable lower exoskeleton system for augmentation of human walking ability, which incorporates human as the integral part of the control system and can relieve human physical fatigue caused by excessive walking and heavy payloads. The contributions of this work include:

1. A new solution to the control scheme of a lower extremity exoskeleton

The methodology of designing an anthropomorphic and adaptable exoskeleton system is discussed. Several control methods of existing exoskeleton systems are reviewed, especially those of HAL-3 and BLEEX.

Different from HAL-3 and BLEEX, the exoskeleton system in this work consists of an inner exoskeleton and an outer exoskeleton. The inner one measures the movements of the pilot, then controls the outer one which supports the payload to walk following the pilot. A special footpad is designed, which can measure the ZMP of the human as well as that of the exoskeleton in real time. Using the measured human ZMP as the reference, the exoskeleton's ZMP is controlled by trunk compensation so that the exoskeleton can walk stably. The human-machine-interface is friendly and no big inconvenience is caused to the users.

2. Development of virtual simulators

It is necessary to verify a control algorithm using a software simulator before it is put into practice, because unsuccessful control may cause great damages to the user or/and

CHAPTER 8. CONCLUSIONS AND RECOMMENDATIONS

the exoskeleton. In this work, an exoskeleton simulator is developed using ADAMS and MATLAB. The simulator is effective and convenient to verify various control algorithms. This simulator can be further used in developing new biped systems or improve the current exoskeleton system.

3. Design and Development of an exoskeleton prototype

This thesis discusses in detail the design and construction of the prototype of the exoskeleton system. The function of each part is explained clearly and various solutions are discussed and compared. Then, the most suitable solutions are selected and the prototype is fabricated and assembled. The prototype consists of seven positive DoF and four passive DoF. It is modular, low cost and adjustable for different users.

4. Implementation of the control algorithm on the prototype constructed

In this work, xPC Target, together with other toolboxes from MATLAB have been used so as to provide a real-time operating system and an integrated development environment for controlling the exoskeleton. Real-time control of the exoskeleton is implemented in this environment. Walking experiments have been performed and demonstrated, which show that the proposed control scheme is workable and the developed exoskeleton can walk stably with the user.

This research is funded by Singapore Ministry of Defence. The report on this work has been submitted to Defence Science Organization (DSO) and the prototype of the exoskeleton system has passed the examination.

8.2 Future Work

The first prototype is only a testbed to verify the control algorithms. The prototype looks bulky and rough. Though the exoskeleton constructed has been verified that it can walk, some topics need to be improved before one will see a reliable and useful exoskeleton that can effectively enhance humans strength and endurance.

Due to the budget limits, we employed existing motors on the previous NTU robotics project to actuate the OLE. Those motors are bulky. Besides, they need cables to supply the power, which limit the motion range of the OLE. In the future, more powerful and compact actuators can make the OLE more lightweight and compact. An untethered, fieldable,

CHAPTER 8. CONCLUSIONS AND RECOMMENDATIONS

and mountable power generation subsystem remains a challenging issue for implementation. Similarly, in the current version, sensors and amplifiers are connected to a desktop computer via wires, which also limit the motion range of the OLE. A portable and independent embedded controller should be built into the OLE and wireless technologies should be applied to the OLE for practical use.

Another area of improvement may be the addition of more passive joints to the OLE prototype. These passive joints may be spring-loaded so that they allow for passive movements without affecting the structural characteristics of the OLE. Examples of passive joints can be at positions like the feet where the passive joint can allow the bending of the OLE foot when walking. This will pertain to the user's extension of the toes while walking. This addition of passive joints at the OLE feet will enable the OLE to be able to reciprocate the human movements better and at the same time, allow the user to have a more natural walking gait while wearing the OLE. Other areas where passive joints can be added are the hip joints and the ankle joints to allow them rotate.

Now it is the inner exoskeleton who senses the user movements. However, the inner exoskeleton increases the distance between the user and the OLE. They can be replaced with goniometers¹, which are more compact and lighter, as shown in Figure 8.1.

Besides, human joints are not simple pin joints but complex joints with a changing instantaneous center of rotation. Compared to the hard linkages of the inner exoskeleton, a goniometer has a telescopic end block that compensates for changes in distance between the two mounting points as the limb moves. The gauge mechanism allows for more accurate measurement of polycentric joints.

To save energy and consequently reduce the size of the power source (e.g. batteries) carried by the OLE, ways to help decrease the torque required at the OLE joints could be developed. Such torque compensation mechanisms are not easy to construct, especially on the OLE. Issues like the amount of torque that will be reduced versus the weight increased by the addition of such a mechanism are widely debatable. Also, such a mechanism should not hinder the movements of the user or the OLE.

Control issues pose another challenging problem for the LEE. With the addition of such torque compensation mechanisms, controlling of the actuators will not be as straightforward. This is because of the additional parameters to controlling the torque output of

¹BIOPAC systems, inc., Dec. 2005, <http://biopac.com/frprod.htm>

CHAPTER 8. CONCLUSIONS AND RECOMMENDATIONS

Figure 8.1: Goniometer attached to the knee joint

the actuators. Nevertheless, a few torque compensation mechanisms will be discussed here, which represents a possible solution for the OLE. Firstly, in the development of the Saika-4 by Tohoku University [85], a method of torque compensation was mentioned. They termed it the “gravity compensation mechanism”. Using a stainless-steel wire and a contrived spring layout, the mechanism is able to compensate without depending on joint angles. This is important, as methods that depend on joint angles compensate with changing forces at every joint angle position. This increases the fuzziness of the control of the system. The mechanism in Saika-4 is reported to be able to reduce the torque requirements of the motors at the robot joints because the robot legs always tend to return to the initial standing position by the spring force provided in the mechanism.

Another method for torque compensation is the application of constant force springs². For example, constant force springs can be attached to the upper limb at one end and the lower limb at the other end such that a constant force is applied between the two limbs. This will result in the limbs having a tendency to straighten.

In conclusion, the OLE prototype acts as a conceptualization of the lower extremity exoskeleton system; future development can be built on this current design. And hopefully, as technology improves, one day, the likes of the LEE will become a part of our normal lives.

²Associated spring Raymond, Apr. 2005, <http://www.raymondasia.com/catalog.htm>

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2. K. H. Low, Xiaopeng Liu and Haoyong Yu, “ Development of NTU Wearable Exoskeleton System For Assistive Technologies,” *Journal of Applied Bionics and Biomechanics*, 3(3), UK, 2006.
3. Leong-Yeo Wai, Xiaopeng Liu, John Heng, K. H. Low and Qing Li, “SEMG Based Wearable Lower Extremity Exoskeleton As an Assistive Technology,” Proceedings of the Ninth International Conference on Mechatronics Technology (ICMT2005), Kuala Lumpur, Malaysia, December 5-8, 2005.
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