Development of Body Weight Support System for Gait Training: Design and Evaluation of the Novel Body Weight Support System Using Pneumatic Muscle Actuators.

A DISSERTATION SUBMITTED TO THE
GRADUATE SCHOOL OF ENGINEERING AND SCIENCE OF
SHIBAURA INSTITUTE OF TECHNOLOGY

by

TRAN VAN THUC

IN PARTIAL FULFILLMENT OF THE REQUIREMENTS
FOR THE DEGREE OF

DOCTOR OF ENGINEERING

SEPTEMBER 2017

To my little family: my wife, Hoang Thi Thu,

and my little daughter, Tran Huong Giang.

Acknowledgments

First of all, I would like to express my special gratitude to my supervisor, Professor Shin-ichiroh Yamamoto who always trusts and supports me in the whole progress of this course. I have learned from him a lot of new knowledge and, especially, the professional attitude of working. Thanks to his kind support and informative comments, I could finish my Ph.D. dissertation. I am also grateful to every committee members, Prof. Akihiko Hanafusa, Assoc.Prof. Nobuo Watanabe, Prof. Masahiro Shibata, and Prof. Kengo Ohnishi, for their patience and many valuable comments such that I could improve my thesis and overcome numerous obstacles facing in this research.

Secondly, I would like to thank the faculty and staff members of Shibaura Institute of Technology for their support. Thank all the NeuroRehabilitation Laboratory members who are always available to help me both in academic studying and daily life in Japan. I will keep in mind every stimulating discussion, the sleepless nights were working before the deadline, and a lot of fun we had in last four years.

Nevertheless, I would like to thank my friends for accepting nothing less than excellence from me. I am appreciated every valuable discussion which helps me a lot in every challenge during my course. Last but not the least, I would like to thank my lovely wife for her sacrifice such that I can keep my mind in my research; she is always behind me, and let me encouragement in each difficult time of my life.

Saitama, Sept 5, 2017

Tran Van Thuc

Abstract

This research thesis introduces the development of Treadmill Body weight support using pneumatic actuator for gait training system. The whole gait training system is so called AIRGAIT system which is for lower limb training for disability patients such as stroke and spinal cord injury (SCI) patients. The main scope of this research is to give a new designing, validating and assessment for active Body Weight support system to reproduce, in the best way, the behavior of the normal walk. Based on the assessments and its evaluations, the novel Treadmill Body Weight support system using pneumatic muscle actuators shows its characteristics of this system are simplicity, low cost, flexibly maintaining the constant unloading force and easiness for controlling the supported force. Especially, the capability of the novel Treadmill Body Weight support system is to generate the unloading forces that track the Center of Pressure (COP) since it switches from left to right and vice versa. Author implemented and compared two different systems, a classic one and new one based on the utilization of PAMs. In the new active BWS system, ie introduced new way so that author used the projection along the x-axis (lateral) of the subject COP as an input in order to give to the patient the freedom from oscillate as during a normal walk. To show the goodness of my proposed active system, several experiments were conducted. For these experiments, author recorded the COP paths of the subject in the implemented systems, the ground reaction force and unloading force of each system for comparison. For stronger system assessment, the Mac3D motion capture system was used for measurements. The Center of Mass and gait parameters for each trial then was calculated

to see the effect of each system to the subject. The COP results show that the active Treadmill Body Weight Support system was the better in reproducing the behavior of normal walking. The same results could be seen from the reaction forces, unloading forces and Center of Mass information. Moreover, the advantage of the new system could easily archive the desired unloading force or reaction force while for the classic one in some case couldn't.

Contents

List of Figures

List of Tables

Chapter 1

Introduction

In two recent decades, almost of locomotion gait training systems has used Body Weight Support (BWS) which enable patients, who cannot fully weight bearing by hemiplegia, paraplegia, or recovering from spinal cord injury or after stroke, stepping in the locomotion walking training system. Conventional therapy treatments even though focused on three aspects muscle activation, weight-bearing capacities and balance [22]. However, despite the muscle activation improvement, these methods did not get fully success gait recovery as well as other aspects of the locomotion gait. The early idea using the treadmill BWS system for human gait training base on the observations from the recovery of spinal cord cutting cats when implementation gait training for them with weight-support and full weight [16] [14]. With the same idea but extending for human gait training, Finch et al., studied the effects of creased weight on electromyography (EMG), kinematic and gait parameters using different levels of unloading load comparing with normal gait [22]. Their findings showed that the Treadmill BWS system expressed the advantaged behaviors in every examined aspect and had a potential extensive applicability in clinical gait training. Martha Visintin et al. examined the differences between BWS and full weight bearing in gait training [26] That experiment was implemented in huge extent by six weeks' gait training with 100 stroke-patient participations in which 50 patients walked under 40 % of their body weight while the rest walk in full weight. The results confirmed that training for a patient who was supported by body weight provided walking ability better than training for who bearing full weight. In comparison with the conventional therapies and gait training using Treadmill BWS, the clinical results showed that therapy with Treadmill BWS system was more advantaged than the old methods because walking with weight support allowed the patient walking with their desired gait [33]. Other tests were implemented for comparison of BWS method versus conventional methods for stroke patients [26][27][1], spinal cord injury patients [19][5][28]. Many researchers investigated a lot of aspects and parameters influence to training, such as age [3][2], the ground reaction force, the symmetry of gait locus [29][1], and the influence of the degree of weight support to hip, knee and ankle parameters [1][29][32][8]. Almost authors agreed that Treadmill BWS system has many advantages in not only for spinal cord injury patient but also stroke, Parkinson and gait recovery for elderly patients. Recent years, Treadmill BWS system in conjugation with orthosis robots were applied for many gait training systems to reduce the therapy cost and labor, for example, WAD system [13], DGO system [36], Lokomat [37] [35].

There are some reasons so that the body weight unloading system plays an important role in gait training system. The first reason is by reducing the gravitational forces acting on the legs by BWS system would reduce the load needs to be overcome by the patient and then the patient could be stepping movements. The other reason is the use of BWS system may be beneficial, because of its dynamic characteristic and special task activity which allow the patient to initiate gait training activities in early after injury [22]. Finally, the BWS system also provided the safety conditions and stability for gait training process.

The gait training processing for incomplete patient normally undergoes many periods depending on the situation of the patient. In the beginning, BWS system often carries out the high level of unloading force during training which to reduce the gravitational force act on the legs of the patient during gait training. The level of unloading force is up to 80% support for body weight because of the weakness of ability bearing for patient's body. The level of unloading force would be reduced following the improvement of the mobility of the patient. The amount of unloading force is adjusted depending on the clinical processing of therapist. In general, the amount of supported weight will be gradually reduced, and the afferent signals to the brain will be increased. In this way, the sensory receptors are activated and improved their functions during the gait cycle.

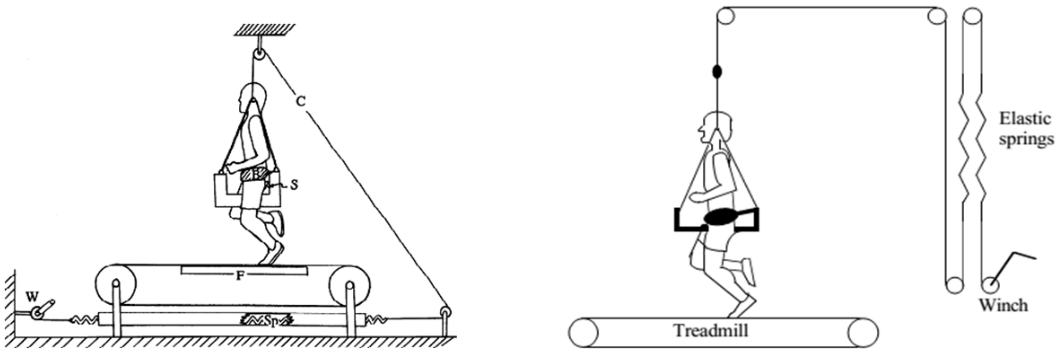


Figure 1.1: Scheme of BWS apparatus using series of springs [30] [18].

Normally, the BWS normally consist of a harness, cables pulleys, frame, and the actuators that carry a percentage of the body weight of the patient. The actuators for BWS system have been developing in two decades, and these were extremely generous. Passive BWS systems, which used a series of springs, were illustrated in the 90s [18] [30]. This apparatus applied a nearly constant vertical force to the body. In this actuator, the limitation can be clearly seen that it may not ensure the unloading force and also, there is not any feedback construction for the unloading force; the amount of force tuning is only by manually. In 2000, a pneumatic BWS system was illustrated by [10], this system concluded a pneumatic cylinder attached to a cart which could be rolling along a track mounted on the frame. The unloading force value can be feedback to the controller, but the system did not control the dynamic force. Additionally, the disadvantage of this system is the cumbersomeness. The dynamic unloading force due to the vertical moving of the center of mass (COM) is much more regardful with time. In 2006, the Lokolift, a fully computer-controller partial BWS system, was developed by [10] which used the dynamic roller combining with parallel spring, could be fully reduced the dynamic unloading force. In contrary, the authors also admit that the physiological gait kinematics and ground reactions together with afferent feedback play an important role in the success of gait rehabilitation, so that the constant unloading of body weight may be not the best solution. In 2010, another Partial BWS system, which characterizes the ability to provide either a constant or synchronously modulated support force, is published by [25]. This system used a linear motor to reduce the gravitational force on the subject body, but this system so cumbersome and not friendly to mount on the gait training system.

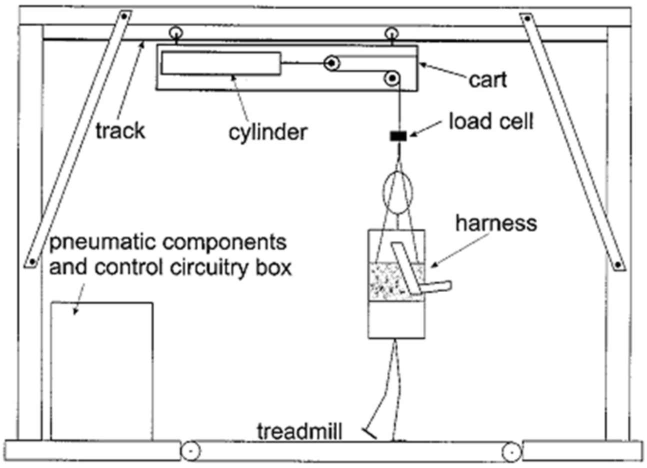


Figure 1.2: Scheme of the WARD system [10].

In short, Treadmill BWS system for human gait training could be classified into three types: Static, passive, and active system [24]. Static systems often consisted of a winch system connected to harness through the rope-pulley. Winch system, attached to a hard frame, could be driven to adjust unloading force by manually wound winch crank or using an electric motor with speed changing by manual [17]. Static systems considered to get constancy weight-support level, however, due to moving in vertical of COM this feature cannot be fit. Passive systems could be applied some kinds of actuators, for example, counterweight [13], extension spring [18], pneumatic cylinder [10]. In these systems, the structures were similar to the static system but the actuators to adjust unloading force were changed. For counterweight system, the unloading force depended on gravitation and inertia of counter weights. In the system using extension spring, the actuator was constituted some springs connecting in series. The passive system also considers to maintain the unloading force as a constant; their characteristics were more advantaged than the static system at which suspension force passively

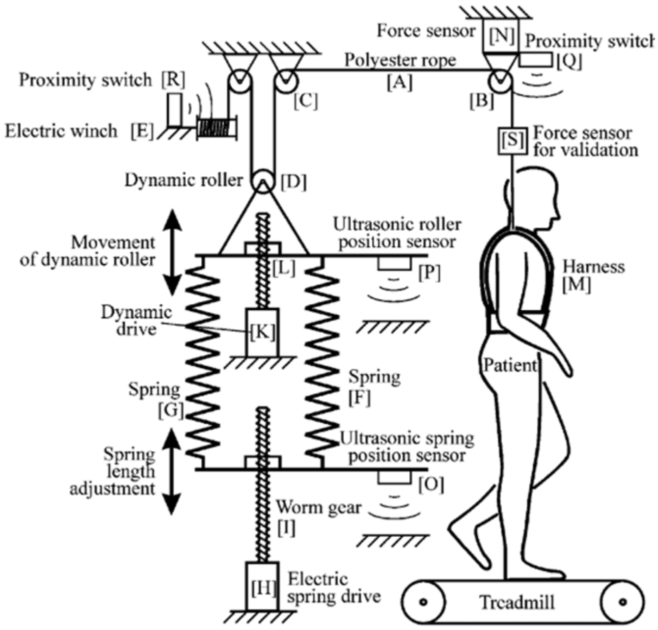


Figure 1.3: The novel mechatronic BWS system using the complicated rope-pulley-spring system [24].

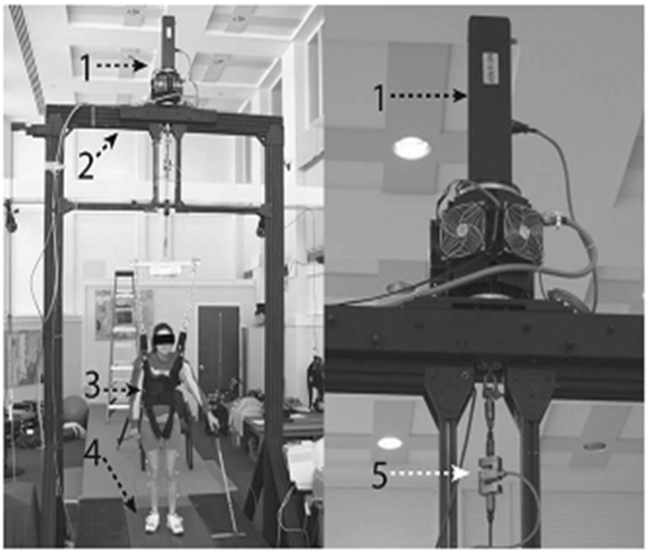


Figure 1.4: A BWS system using electrical motor [29].

adapted, but uncontrollable, with moving of COM during walking. The adapted suspension force is a potential force such as gravitation, elastic force or displacement of the cart in the case of using pneumatic cylinder. Limitations of passive systems can be seen that they could not guarantee constant force, and by using a rope-pulley system, the subject could be swung during walking. In contrast, active systems could overcome almost of the limitations of passive systems; they could be sure constant weight support by combining close-loop control system with dynamic roller [24] [23] or electromagnetic motor [29]. Also, Franz et al., suggested that unloading force which was a variation of gait pattern had more efficient than the constant force. However, almost active systems, which were applied the inherent structure in which actuators were connected with subject trunk through a pulley-rope mechanism, had a disadvantage that subject using these systems would be swung. The critical point of gait training is that center of human body always changes in three-dimension space. Franz et al. suggested that synchronizing between gait and force modulation during stance phase could provide more efficient in gait training [29]. However, no study successes modulate the support force in other dimensions of moving COM as well as consider the movement of the center of pressure (COP).

1.1 Problem Statement

This research introduces the development of Treadmill BWS using pneumatic actuator for gait training system. The whole gait training system is called AIR-GAIT system which targets lower limb training for patients with disability such as stroke and spinal cord injury (SCI) patients. Based on the assessments and its evaluations, the novel Treadmill BWS system shows its characteristics of this system are simplicity, low cost, maintaining the constant unloading force and easiness for controlling the supported force. Especially, the capability of the novel Treadmill BWS system is to generate the unloading forces that track the COP (COP) since it switches from left to right and vice versa. To the authors' best knowledge, BWS system that emphasizes on the control unloading force which tracks COP using Pneumatic muscle actuators in the gait rehabilitation field is not yet to be extensively investigated and commercially available. Also, as mention above, even though there are numerous of BWS system has been developed, the limitations of rope-pulley type are the swinging of COM and difficulty of modulating the unloading force follow the moving of COP during patient gait training. Then authors' motivation is to design and develop a very simple Body Weight unloading system with its capabilities generating active unloading forces and tracking the COP moving. This thesis focuses on widely activities from system design, control and system assessment for the new system including comparisons the new Treadmill BWS system and the Counter Weight system (the previous one).

1.2 Objectives

This research embarks on the following objectives:

- **i**. To design the novel BWS system for gait training and implemented in the AIRGAIT system by using pneumatic artificial muscles.
- **ii**. To derive and design and validate control scheme for the BWS system using pneumatic muscles base on real time control system and FPGA platform.
- **iii**. To evaluate the effects of the new strategy implement unloading force based on the new BWS system using motion capture system.

1.3 New Findings/Knowledge

This research will result in new Body Weight unload system and a potential pilot strategy unloading force for locomotion rehabilitation system. Concept and knowledge from this research could lead to new explorations as below:

- **i**. Exploration of the Treadmill BWS system which uses pneumatic muscle very similar to human muscle as lifting actuators.

- **ii**. Derivation of the control strategy and a novel unloading force generated that enables to follow the moving of COP during gait locomotion.
- **iii**. The new strategy using the Treadmill BWS system for gait training system as well as the alternative for clinical rehabilitation training can be evaluated. The flexible providing unloading force of the new BWS sytem could provide very flexible gait training strategy in clinical practice.

1.4 Significance of Research

Because of increasing number of elderly persons in developed country and growing number of accidents in developing country, the human society requires more and more the medical and rehabilitation devices. The conventional therapies, even current treatments, needs much labors and efforts for rehabilitation especially gait training for the patients who are recovering from spinal cord injury or stroke and patient who need locomotion training for the lower limbs. The conventional therapy programs necessitate the intensive efforts of two or event three therapist to treat only one patient standing and walking step by step slowly. These methods could highly and potentially cause fatigue for both patient and therapist, and the rehabilitation goal was hard to control and assessment. Nowadays, modern rehabilitation approaches applies many technologies to help reduce the labor cost and totally improve rehabilitation results. The prime of technology implemented in rehabilitation is BWS system because of two reasons. The first reason is the BWS system provided a safety apparatus for gait training, the second goal of using Body Weigh support system is to help the patient to reduce the force needed during gait training then reduce the therapies labor cost. Developing of technology can reduce much effort of physiotherapy, for example, people could apply many kinds of mechanisms for BWS system as well development of leg robotic orthosis. Therefore, the new rehabilitation device became an effective tool for the therapist to apply many lessons on gait training to gain the best efficiency, in this case, beside of leg robotic orthosis the BWS system also plays a more important role.

From the authors observation, intuitively, the rehabilitation device cost is a big problem in the developed country in general as well as for individuals. The lower limb gait training system could be only approached by some big hospital or national rehabilitation center; this then leads to high rehabilitation cost which prevents lower income people could be used, especially in developing countries or develop countries. Another problem is that the expensiveness of rehabilitation devices will prevent their spreading and also home therapy. If the patient could begin therapy sessions quickly, he/she could take a shorter time for recovering. Unfortunately, the more technological and efficient rehabilitation device is the higher cost of the instrument is. It is because the advanced gait training device has been used much expensive and high precision apparatus like electric servo-motor, linearity actuators, and other sophisticated mechanisms. The using of Pneumatic muscle in rehabilitation system is an indisputably competitive advantage in comparison with the other apparatuses. Moreover, this actuator could be applied to many rehabilitation devices not only to lower limb rehabilitation device but also to the upper limb rehabilitation device. This kind of actuator could reduce the cost for the health-care instruments which have big meaning for spreading the rehabilitation device and also for developing countries.

Moreover, the development of BWS systems based on the pulley-rope mechanism in recent years were aimed mainly to get the constant unloading force. This could be difficult if therapist wished to change the strategy of the gait training lesson, for example, applying unloading force base one COP. In fact, almost current BWS systems were applied the inherent structure in which actuators were connected with subject trunk through a pulley-rope mechanism. Therefore, this structure could have a disadvantage such that subject using these systems would be swung during walking. We should know that one of the critical points in gait training is that the center of human body always changes in three-dimension space. This research will highly introduce a different BWS system in which the novel structure applied very flexible actuators. The AIRGAIT system using the novel BWS system could allow patient could walk in not only different of unloading force level but also even applied unloading force flow the moving of COP. Furthermore, the leg robotic orthoses integrated which also used pneumatic muscle allow a patient who disables his/her legs could repetitively follow gait pattern at different speed. The measurement system integrated is to identify the subject's walking parameters like COM and COP. This information will help therapists analyze the health condition of the subject, and then therapist could have specific treatment strategy depending on the level of training.

1.5 State of the Art

In this research, the author proposes an entirely new way to think the rehabilitation of patient on the treadmill. The author is strongly convinced that the common static BWS systems are not comfortable enough to give good results on the rehabilitation of patients. The main scope of this research is to provide a new designing, validating and assessment for active BWS system to reproduce, in the best way, the behavior of the usual walk. We implemented and compared two different systems, a conventional and new method based on the utilization of PAMs. The main differences between the systems based on PAMs are just under the control of the unloading force. In the active BWS system, the author uses the projection along the frontal axis of the subject's center of pressure (COP) as an input to give to the patient the freedom to oscillate as during a normal walk. To show the goodness of the proposed active BWS system, author recorded the reaction force, COP path, and COM trajectories of the subject in the implemented systems. The results show that the new active BWS system is the best in reproducing the behavior of a normal walk. Future work will focus on giving more freedom to the patient especially by also considering the oscillation on the sagittal axis.

1.6 Scopes and Limitations

- **i**. The designing of Treadmill BWS system based on developing pneumatic muscle and the frame that integrated into a bigger gait training system - the AIRGAIT system which is developing in our laboratory.

- **ii**. All the measurements, control system, experimental tests, and design will be based on the developed BWS gait training system of AIRGAIT exoskeleton.
- **iii**. The simulation and control design are coded in MATLAB language, while the control system programs are coded using LabVIEW real-time and LabVIEW FPGA. The measurement system is based on Cortex and Motion capture system.

1.7 Outline of the Thesis

The title of this research is " Development of BWS system for gait training: design and evaluation the novel BWS system using pneumatic muscle actuators." This section briefly describes the content of the research thesis which consists of six different chapters including introduction, Pneumatic Muscle Actuators Characterization, mechanical system design, control system, results and discussion, and conclusions.

- **Chapter 1**: The first chapter provides a general introduction, overview, and background of the whole research including the problem of the statement, specific objectives, scopes and limitation, and outline of the thesis.
- **Chapter 2**: The second chapter is to describe in detailed the Pneumatic Muscle Characterizations of Pneumatic Muscle Actuator which is developed by our laboratory.
- **Chapter 3**: The third chapter is the design system and evaluation section for the Treadmill BWS system using Pneumatic Muscle Actuators. All of the mechanical were described thoroughly in this section.
- **Chapter 4**: The fourth chapter describes the control system development for the novel Treadmill BWS system. In this section, COP tracking model scheme and strategy were introduced. All the kinematics analysis and the mathematical derivation of the COP tracking model that generates the

input reference for the actuators were described in details. Furthermore, the controller design and the control implementation base on LabVIEW real-time and LabVIEW FPGA were also described.

- **Chapter 5**: The fifth chapter consists of the results and discussion of this research project. The first sub-topic would describe the experiment design for system validation. After that, several sub-topics delves into the assessments evaluation results and analyzing. In this chapter, the validation experiments were designed including BWS system and Counter Weight system for comparison. The motion capture system with six cameras also was used for recording twelve reflexed markers to calculate COM for comparing between the new system and the old one. Moreover, in the experiment assessment for the new BWS system, unloading forces, reaction forces, COP (COP) also were recorded at several weight unloading levels for both BWS system and Counter Weight system so that we could have the whole picture of the advantages and disadvantages of the new system.

- **Chapter 6**: The last chapter deliveries the conclusion of the entire research's assessments and provide some recommendations for an upcoming project which keeps continuing to improve the system such as in design, control system. The limitations of this research project also are delivered in this chapter.

Chapter 2

Characterization of the Pneumatic Muscle Actuator

The applied Pneumatic Muscle Actuator with 1.0" [inch] diameter in this work was McKibben braid pneumatic actuator type that has been developed in our laboratory as shown in Figure 2.1. An internal bladder tube of Pneumatic Muscle Actuator was made from rubber and was surrounded by braid mesh. Both ends of the bladder and surrounding layer were attached to the fitting. The input air pressure was separately supplied for each PAM and regulated by single pressure regulator. When PAM is provided pressure with input signal, it would contract and generate force like a bio-muscle.

Two experiment were conducted to investigate the characterization of Pneumatic Muscle Actuator: the first experiment is to see the static characterization, and the other one is to understand the dynamic behavior of the Pneumatic muscle Actuator.

Figure 2.1: Pneumatic artificial muscle (PAM) – McKibben.

2.1 The static Pneumatic Muscle Actuator Characterization

Figure 2.2 shows the experimental set up for conducting the statics characterization. The aim of this experiment is to see the maximum responded force for a specific pneumatic muscle. A characterization of pneumatic artificial muscle was implemented to investigate the behavior of the actuators. The experiment was set up such that both tips of PAM were fixed in a solid frame which has length equals to the free length of PAM; then the pressure was supplied into PAM in equidistant levels of 0.05 MPa and from 0.05 to 0.5 MPa. The measurement system records the output forces for each level of pressure.

Figure 2.3 shows the characterization of the pneumatic muscle actuator. Because the problem of the BWS system is the force control, there force the characteristic of pressure and force in the pneumatic muscle actuator is consider. With data collected from this experiment, a simple model was gained by fitting PAM data with a polynomial function. In this model, the fitting equation is solved in term of pressure. The fitting equation, 2.1, was expressed as a relationship of two variables Y and X instead of respectively force and pressure, presented sensibility analysis was conducted to evaluate the degree of the proper function.

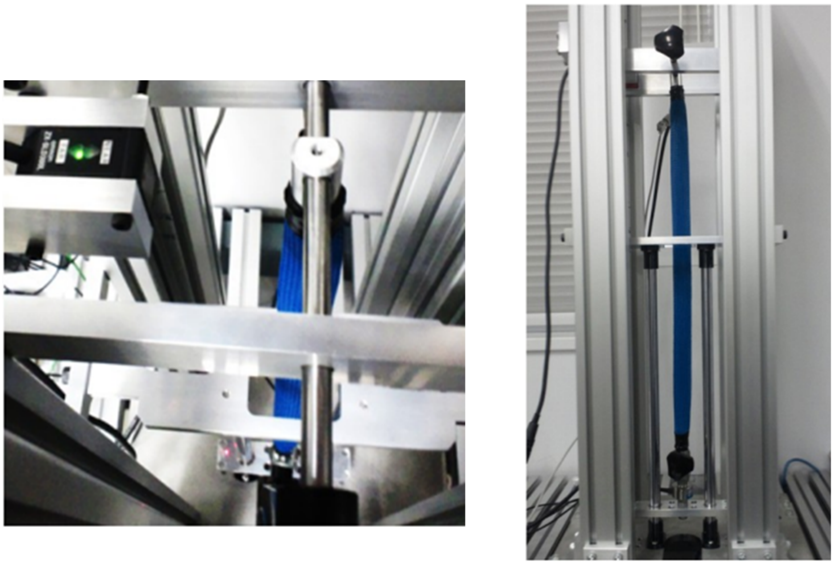


Figure 2.2: The Testbed used for conducted experiments to investigate the characteristic of the pneumatic muscle actuator.

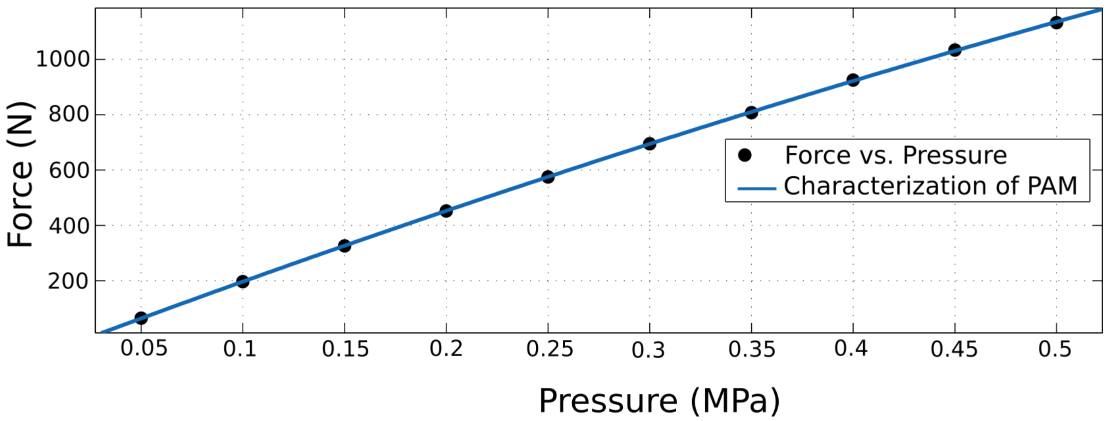


Figure 2.3: Pneumatic muscle characterization: The Force-Pressure relationship.

The Table 2.1 represents the parameters for static force pressure model of the PMA characterization. The Root Mean Square Error between the experimental points and the fit equation is computed and showed in Table 2.2 as a function of the degree of the equation. It is possible to see that the error was small even though in the first order of fitting equation. Then fitting equation was chosen as the first order polynomial equation to express the characterization of PAM in the relationship between force and pressure. The reduction of error would be seen in the second order fitting function but not so much in the higher order of equation.

$$Y = p_0 * X + p_1 \tag{2.1}$$

Table 2.1: Representation of the parameters for static force pressure model of the PMA characterization.

Parameters	p_0	p_1
Value	2447	-56.87

Table 2.2: Sensibility analysis of fitting curve of the experiment data.

Degree	**RMSE**	**R square**
1	14.4371	0.9986
2	2.3617	1.0000
3	2.1342	1.0000
4	1.9637	1.0000

2.2 The dynamic Pneumatic Muscle Actuator Characterization

To investigate the speed of response of the system, we conduct a dynamic characterization of the PAM. Particularly, we fix the PAM in a bench, and we stress it with step functions by supplying it with pressures ranging from 0.05 up to 0.5 MPa with increasing step 0.05 MPa and recording the equivalent force through a load cell. The data from the dynamic characterization of PMA would be consider in the system identification for the control system. The detail of the system identification procedure will be discussed in Chapter 4 The results are shown in Figure 2.4 where we can see that the response of the PMA is very quick and it can be considered appropriate for the specific application. In the Figure 2.5, the dynamic characterization of PMA was depicted clearly. We could observe that the PMA activated after around 100 (ms) and reached to the steady state after 300 (ms). The time delay would affect to the quality of the control system.

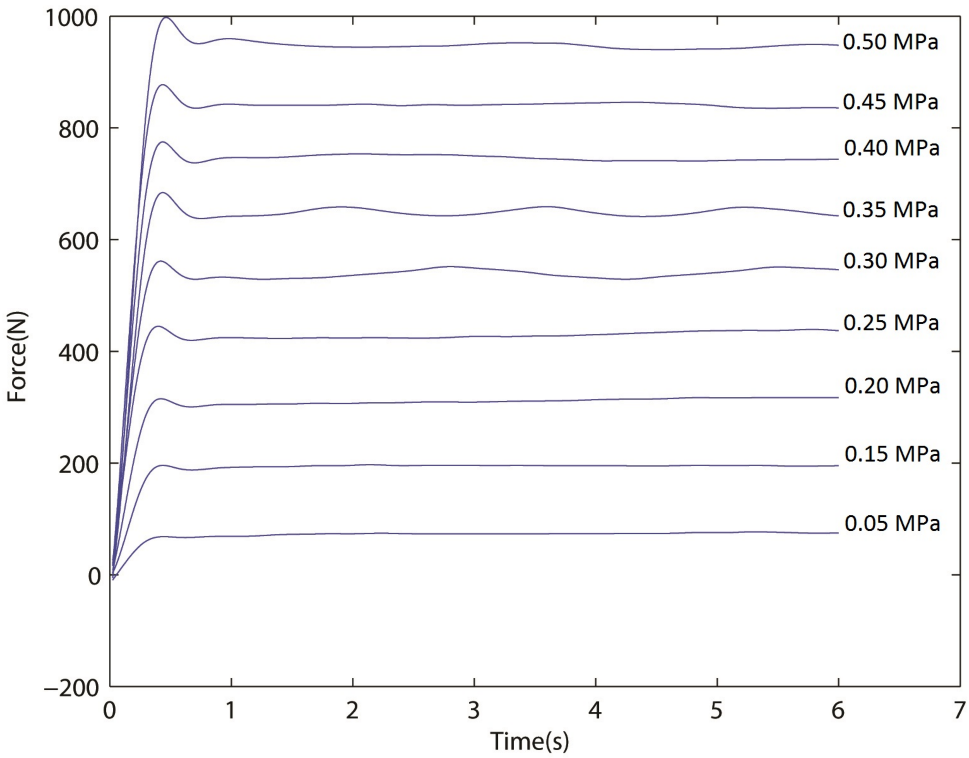


Figure 2.4: Dynamic Characterization of Pneumatic Muscle Actuator.

2.3 Conclusion

This chapter introduced the Pneumatic Muscle Actuator (PMA) that would be applied in the novel BWS system. The characterization of the PMA also has been

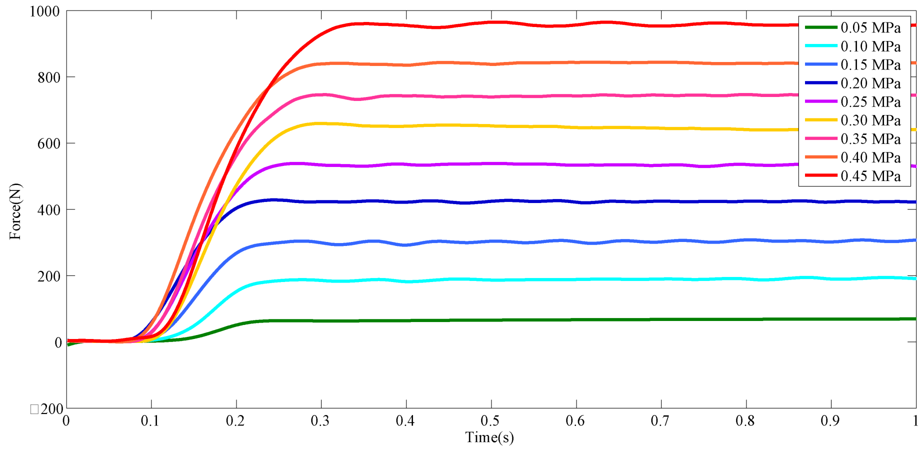


Figure 2.5: Dynamic Characterization of Pneumatic Muscle Actuator: scale at the range of one second.

investigated. In the statistic state, the relationship between the generated force and pressure of PMA is linear. However, in dynamic character, the PMA only activated after a piece of time. This nonlinear behavior of PMA would affect to the quality of the control system.

Chapter 3

Mechanical system design

3.1 Introduction

In the third chapter, author discusses the mechanical system design procedure for the BWS system. The first subsection delves some possible method that would apply for the BWS system. In the second subsection, the mechanical system design of the BWS system is discussed in detail.

BWS system has been developed in more than two recent decade for gait training. The conventional BWS systems commonly applied the inherent mechanical structure in which the actuator was used the rope-pulley mechanism connected to subject trunk. The "pendulum effect" would appear in gait training and made the patient uncomfortable during gait locomotion. In this study, a new BWS system which applies the PMAs is proposed. The new BWS system has a very low cost and simple mechanical system, however, it could overcome every limitation of the previous BWS system.

The Treadmill BWS system and robotic leg orthosis exoskeleton incorporate as two main parts of the AIRGAIT system. Figure 3.1 shows for the overview of the schematic diagram for the BWS system using four pneumatic muscle actuators and the Figure 3.2 shows the mechanical prototype of the BWS system. The BWS system is one of three main part of the AIRGAIT which integrated the robotic leg orthosis system has six PAMs which antagonistically arranged in pairs like the human musculoskeletal system (i.e., mono- and bi-articular muscles), see [11] for more details. The design of the mechanical structure of BWS system was described in the previous paper [34]. Currently, the AIRGAIT system employs the National Instrument device – NI cRIO-9176 which utilizes the NI Control and Simulation Module and LabVIEW My Rio software as the operating system. The level of unloading force is inputted in the program in host computer, then the data is transferred by the local network to the real-time computer which is integrated FPGA chip. Finally, the command is sent from the real-time computer to FPGA by using FPGA interface functions. The signals from load cells are used to calculate the COP tracking model and generate the reference signals. The control signals are outputted from NI output module to operate the electron-pneumatic regulators (ITV 2050 by SMC). In this system, each of muscle is implemented by a regulator. The measurements by the system (i.e. Force (N)) provide the feedback signal to the controller through A/D input modules from National Instrument. The force sensors are given by KYOWA (LUR – A – 2KNSA1). For each pneumatic actuator also has a small pressure sensor for monitor the pressure inside. The four load cells for measurement of the COP moving are integrated type under the motion base of the treadmill. The treadmill is provided by Force Link.

3.2 Selection of Actuator for BWS system

There are many mechanical structures investigated. The original idea is to design a simple BWS system using pulley-rope, springs and an electric motor as shown in Figure 3.3 and Figure 3.4. For the Figure 3.3, the principle scheme shows the system consists of a harness, frame, the pulleys-roller system, winch motor and the actuator to compensate the gravitational force on the patient. The actuator, here, uses some counter weights which are mounted on series with two spring; the other end of springs is mounted on the dynamic pulley. A Pneumatic Cylinder is fixed, but the piston of this cylinder is also mounted on the dynamic pulley. To carry gravitational force of the patient, the therapist must choose some of the counter weight which is equivalent to the percentage of unloading force, while the

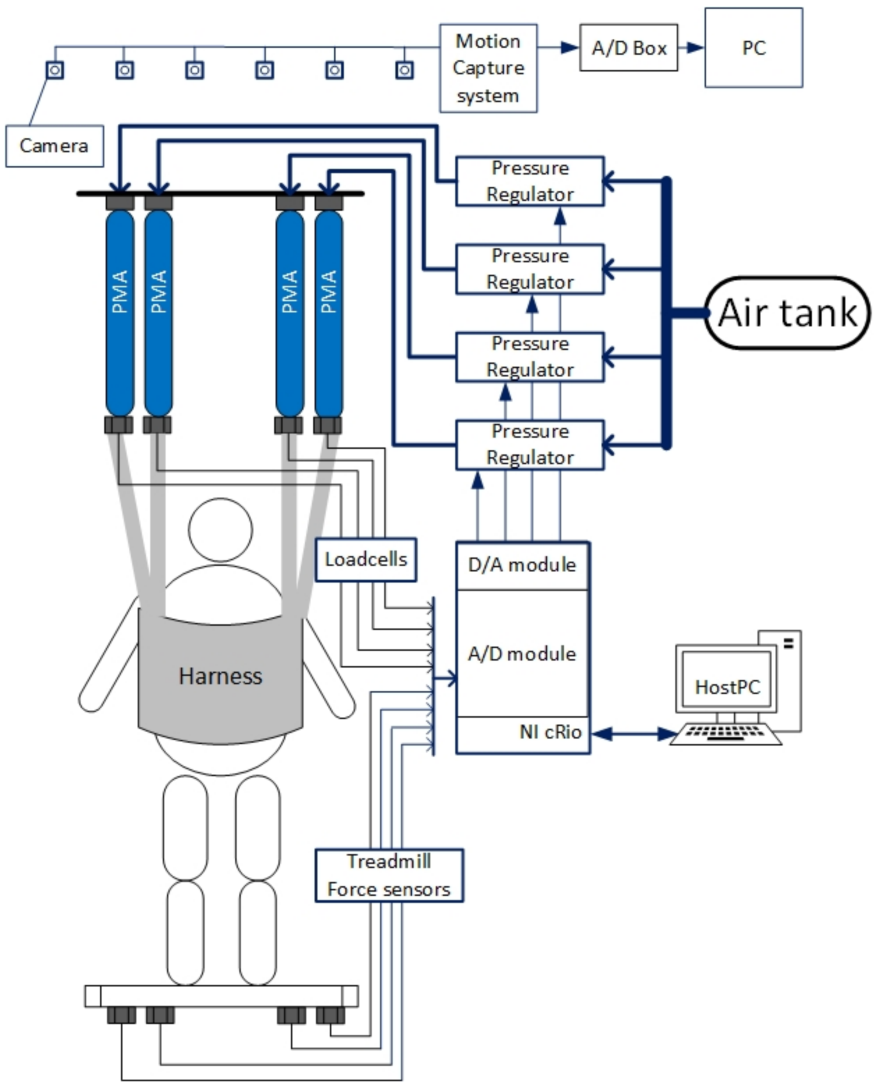


Figure 3.1: Scheme of the BWS system using pneumatic muscle actuators is represented.

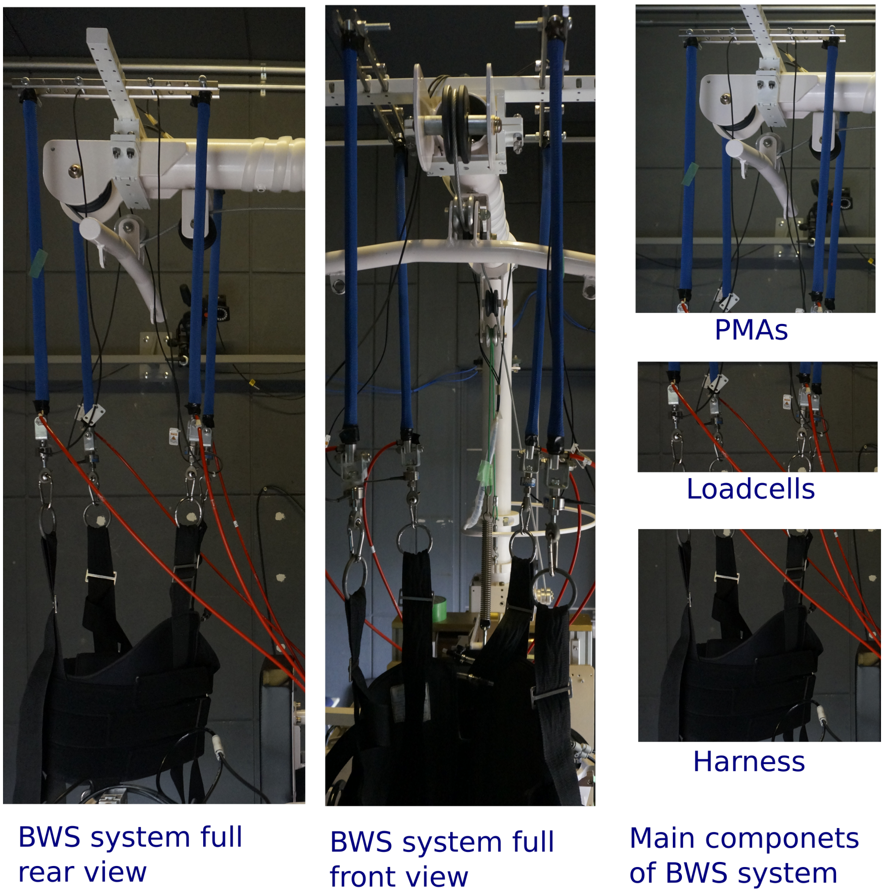


Figure 3.2: The prototype of the new BWS system.

pneumatic cylinder will adjust the dynamic unloading force which is generated by the movement of the COM. This system has the advantage that is the simple structure and can be absorbed vertical force, but it is a little bit difficult to control and maintain constant unloading, it may be not convenient for the operating of the therapist.

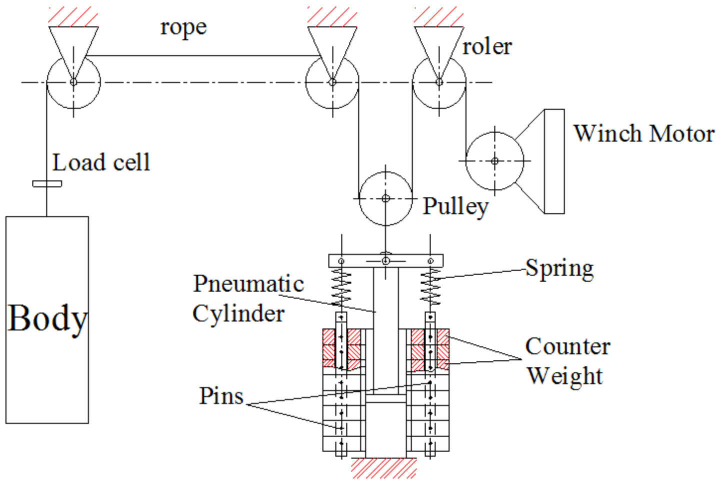


Figure 3.3: Scheme for the first idea of the active BWS system which combines the pulley-rope mechanism and the counter weights.

For the Figure 3.4, The system is quite similar to the first scheme, the second scheme also uses a pulley system combining with an actuator to reduce the gravitational force on the patient, but it has two dynamic pulleys, and the structure of the actuator is very different. The system has some springs which are mounted on a moveable platform; the other ends of springs are attached to the dynamic pulleys. The system also has a fixed pneumatic cylinder, and piston rod of this Pneumatic Cylinder is mounted on the dynamic pulleys. The moveable platform, which is to adjust the deflection of springs so that the unloading force can be changed, is driven by a ball-screw using a servo motor. The pneumatic cylinder bears a part of the unloading force and also adjusts the dynamic force. It can be clearly seen that, due to use two dynamic pulleys in this system, the required unloading force value is four times as high as the amount of patient weight carrying, but the deflection of the dynamic pulley is four times as small as the movement of the patient's COM. This lead to spreading the scale of the

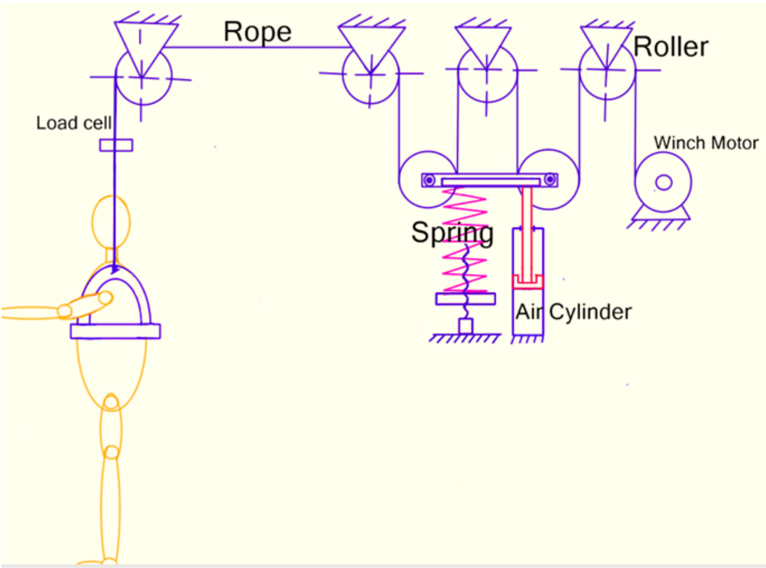


Figure 3.4: Scheme for the second idea the active BWS system which combines the pulley-rope mechanism, spring and air cylinder.

dynamic unloading force which is minuscule, and reducing the oscillation of the counterweight due to the movement of the COM. However, the system is still too complicated because there are many actuators in the scheme and even these actuators are similar capability but are used for the same goal.

Figure 3.5: The linear electric motor.

The other choice for replacing all the complicated actuator in the above scheme is to use electrical linear actuator as Figure 3.5 . The advantaged of this actuator is very compactness and high capacity, the idea is applying this actuator as the main actuator to adjust the unloading force to the requirement level. The limitation of this type of actuator is that when it connects to the pulley rope system, then it would lose the redundant characteristic like spring and difficult to compensate for the dynamic unloading force since the COM suddenly moves up and down.

Some technical information:

- Voltage: 12V DC Linear Actuator.
- Stroke Size: 1- 24" (25.4 – 609 mm)
- Load Capacity: up to 850 lbs (385.55 kg)
- Speed: 0.59"/sec (no load) (14.968 mm/sec)
- Type of duty: 20%)
- Operational temp: -25^{0}C +65^{0}C
- Protection class: IP65
- Certification: CE
- Built-in limit switches, not adjustable.
- Product Weight: 9.55 Lbs (4.332 kg)

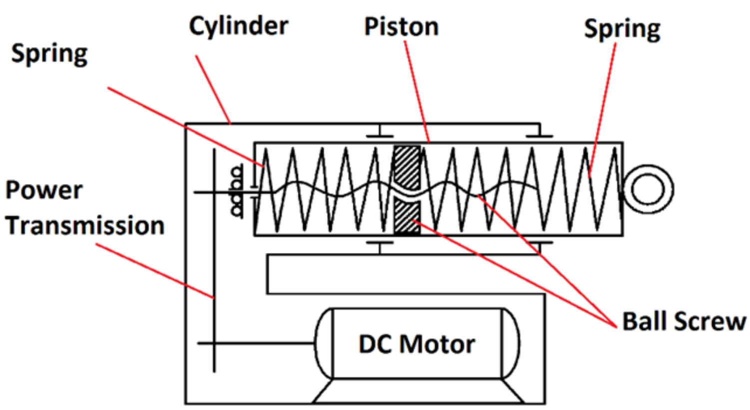


Figure 3.6: Principle scheme of the Linear elastic actuator.

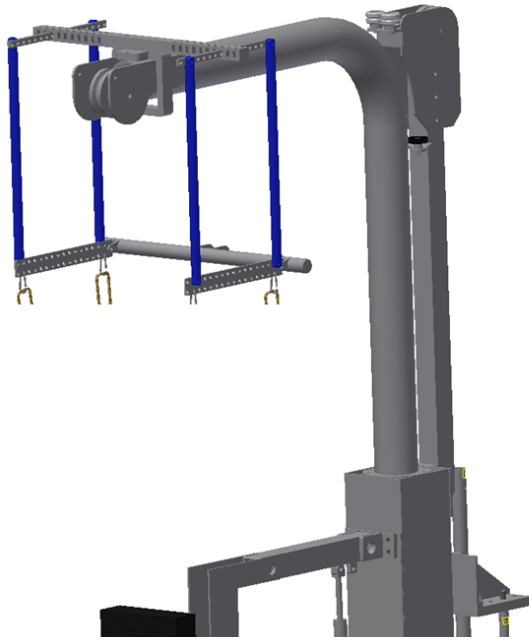


Figure 3.7: The new design for BWS system using four pneumatic muscle actuators.

The elastic actuator was also considered for using as the main actuator to adjust the unloading force. The idea of the elastic actuator using electrical actuator for BWS system with the principle scheme is shown in Figure 3.6. In gait training, one of the most important things we have to consider is the mechanism to conserve the expenditure energy during walking. That means the gravitational potential energy transfers to the kinetic energy due to the moving of the COM during swing phase. So this actuator has an advantage that it will store potential energy in the first stage of swing phase and restore this energy in the rest stage of swing phase due to the spring element. However, this actuator is still difficult for manufacturing, and maybe the price will be very high. Another limitation of this system is that since connecting one tip to the pulley rope system its capacity is only to try to maintain unloading force as constant, it cannot remove the fluctuation due to the pendulum mechanism since a rope connects to the trunk.

Finally, developing a new prototype for BWS which combines four pneumatic muscle actuators is very simple structure and easy to implement in practice as shown in Figure 3.7. Pneumatic muscle is chosen for this BWS system due to its advantages as a high power-weight ratio, high power-to-volume ratio, and inherent compliance. It represents its behavior is very suitable for applying for BWS system since it can be connected directly from base frame to the harness. The pneumatic muscle is very compact, flexible and vigorous; it can easily mount.

3.3 Mechanical system and disposition of Pneumatic muscle

The BWS system consists of a huge inverted L-shape beam. Four Stainless steel bars are separated in two couple, and each of couple is mounted in each left or right side of strong steel bar. Four pneumatic muscle actuators are mounted to the steel frame. Each of these muscle has with one tip fixing on the beam and the free tip connecting to harness. The positions for mounting pneumatic muscle actuator are symmetrically arranged so that the total unloading force generated will be balanced and reduce the displacement of the mounting point as well as the vibration during operation. The designing and dynamic simulation analyzing are implemented by using Inventor software by Autodesk. The detail for the design is introduced in Appendix A. The simulation analyzing is conducted for the most dangerous case since all the load is concentrated at the furthest position as shown in Figure 3.8.

The simulation result indicates that in the most dangerous case the Von Mises street reach to 593 Mpa still smaller than the maximum permission stress for the bar made from stainless steel as shown in Figure 3.9. One of the most important thing in this analyzing is the displacement of the mounting point of muscle. Figure 3.10 shows the simulation result for the most dangerous case; we could see that the maximum displacement reaches to 1.5 mm, this displacement is tiny comparing with the moving of COM and may not effect to the amplitude of the generated unloading force.

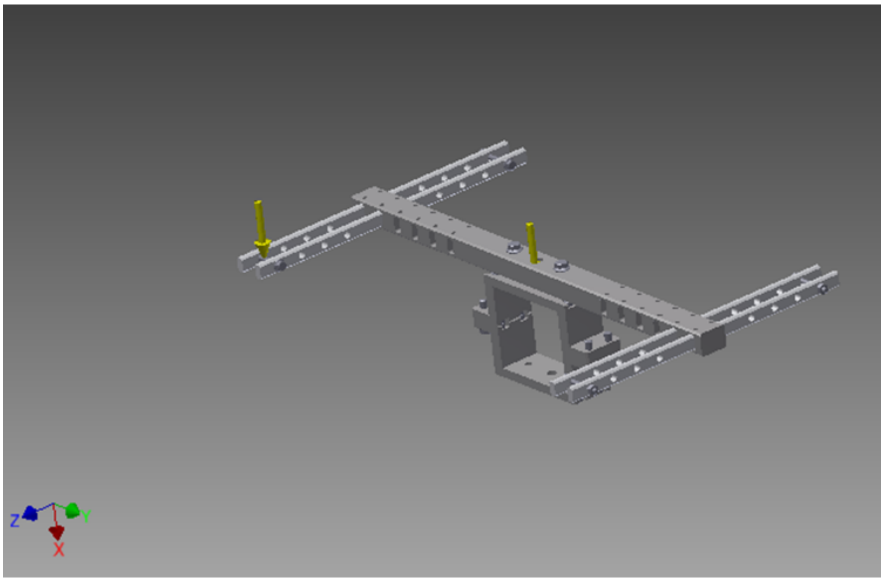


Figure 3.8: The most dangerous case since all load concentrate at the furthest position.

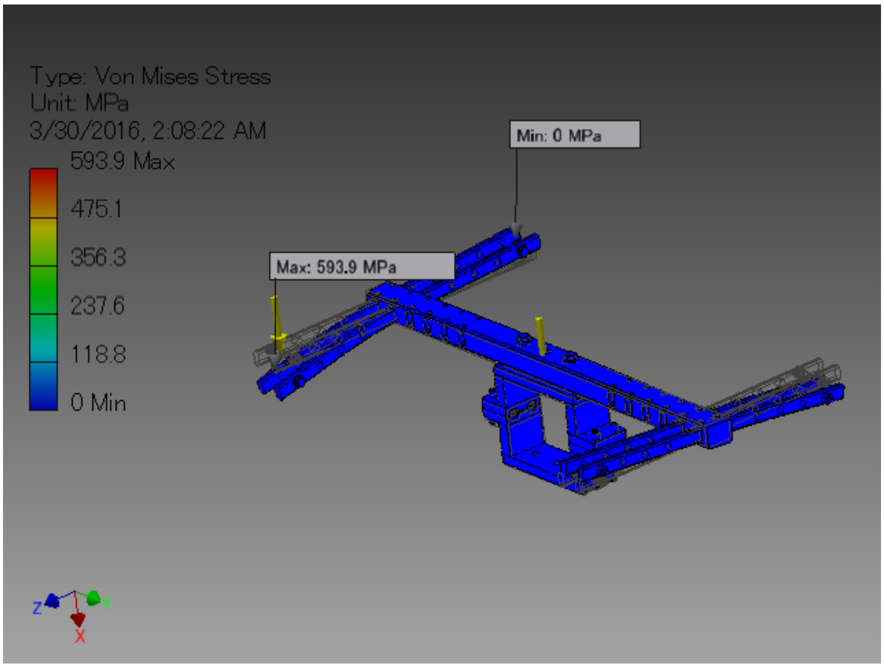


Figure 3.9: The Dynamic Simulation result in which the maximum Von Mises stress reach to 593 Mpa is represented.

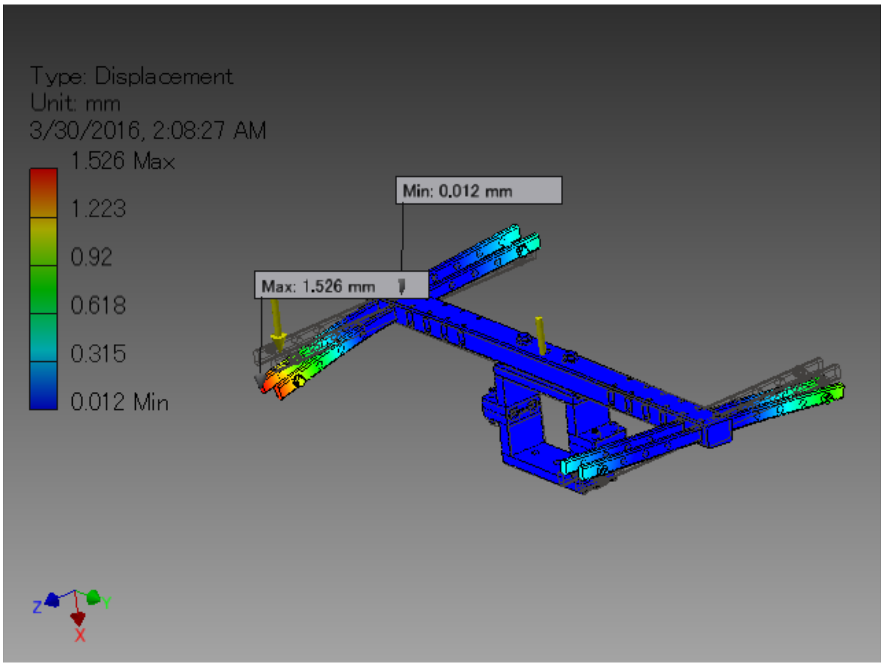


Figure 3.10: The maximum displacement is reached to 1.5 mm.

3.4 Conclusion

This chapter introduced the overview of AIRGAIT system which includes the Treadmill BWS system. This chapter also introduced numerous of methods applying for BWS system and represented the advantage of the BWS system using pneumatic muscle actuators. This chapter also described for the mechanical designing and analyzing the dynamic behavior of the structure.

Chapter 4

Control System

4.1 Overview of the control system for Pneumatic Muscle Actuator

Development of modeling and control system for pneumatic muscle actuator has been considered for two recent decades. Cadwell et, al.. developed a control system for lower limb robot using pneumatic actuator [9]. The simple model for PMA based on the geometry parameters of the actuator. Chou and Hannaford developed a static model for PMA base on the energy conservation during compression and expansion of muscle [7]. The quasi-static and dynamic experiment also represented the hysteresis phenomenon in the relationship between contraction and pressure. This model, in which the thickness and the friction of the air muscle were considered, then also was developed by Tondu and Lopez [6]. Most of the models based on geometrical and physical principle are too complicated. There are many parameters unknown and even impossible for observation. Moreover, the geometry and physic base models did not demonstrate the nonlinear hysteresis phenomenon. Recent years, many models basing on experimental observation has been developed to overcome the limitations of classic models [20] [21]. Maxwell-Slip model is to cover the hysteresis phenomenon of muscle length and pressure. This model was introduced by Tri Vo Minh et, al… based on isometric test and isotonic test [38]. George Andrikopoulos covered Bouc Wen model and PI model [4] [12]. Most of these models tried to interpret the hysteresis model and apply to the very high precise position control system [15] [31]. These models are too difficult to implement in real-time control because of heavy computation for each cycle time. Moreover, in this research, position precise control of PAM is not in scope, instead, force-pressure control is considered because of the characteristic of unloading system which does not require the too accurate value of unloading force. Control design in this chapter is to consider to the simplified linear model so that the controller can reduce the time computing in each cycle time.

4.2 Identification simplified linear model for pneumatic muscle actuator

In this section, a Pneumatic muscle actuator model is obtained by linear identification method. An experiment was set up for dynamic system identification. The experiment setup was similar the experiment for investigation the mechanical properties of the Pneumatic muscle. It means the object muscle was fixed both two tips in nominal length. A step function pressure signal was supplied inside the muscle. Measurement system would collect the force responded by the muscle. After that, the sine wave pressure signals were also provided in several frequencies. The aim of these experiments was to validate the estimated model later. Then, the collected results were analyzed using MATLAB for identifying the model using Linear Model Identification module integrated. For simplify, transfer function model with no zero and two poles was selected.

The continuous linear transfer function is represented by:

$$Y(s) = \frac{b_0}{a_0 + a_1 s + a_2 s^2} * U(s) \tag{4.1}$$

In this equation: Y(s), and U(s) is the output signal and input signal in frequency domain. a_0, a_1, a_2, and b_0 represent for coefficient parameters of the denominator and the numerator polynomial.

Table 4.1: The estimated parameters for the continuous model using Matlab.

Parameters	a_0	a_1	a_2	b_0
Value	116.3	15.95	1.000	27.83

Discrete the derived continuous linear transfer function, from the equation 5.1 we have:

$$Y(z) = \frac{b'_0 + b'_1 z^{-1} + b'_2 z^{-2}}{a'_0 + a'_1 s + a'_2 s^2} * U(z) \tag{4.2}$$

In this model Y(z) and U(z) are discrete output and input signals. $a'_0, a'_1, a'_2, b'_0, b'_1$, and b'_2 represent for coefficient parameters of the denominator and numerator polynomial.

Table 4.2: Representation of the estimated parameter for the discrete model at sample time equal to 0.001 seconds.

Parameters	a'_0	a'_1	a'_2	b'_0	b'_1	b'_2
Value	1.000	-1.984	0.984	0.000	1.3845-05	1.377e-05

Table 4.1 and Table 4.2 represent the parameters estimated for the Pneumatic muscle actuator model since considering to force and pressure. Figure 4.1 shows the fitting of pneumatic muscle model and the measurement data. The calculation showed the fit to estimated data up to 87.7%. The fourth order of transfer function model is also considered, and the data are fitting up to 94%. However, the higher order of the transfer function, the more difficult in control design step and longer time computing in the real-time system. Then, in this study, for simplicity, the lower order of transfer function model for pneumatic muscle was selected.

In Figure 4.2, 4.3 , and 4.4 the comparison between estimated model and measurement data in case of the reference data is a sinusoidal wave inputted at different frequencies. We could see that the model shows its capacity proper fitting with the real system, particularly at the low frequency. At high frequency, the model still shows its quite good fitting with the actual data however the

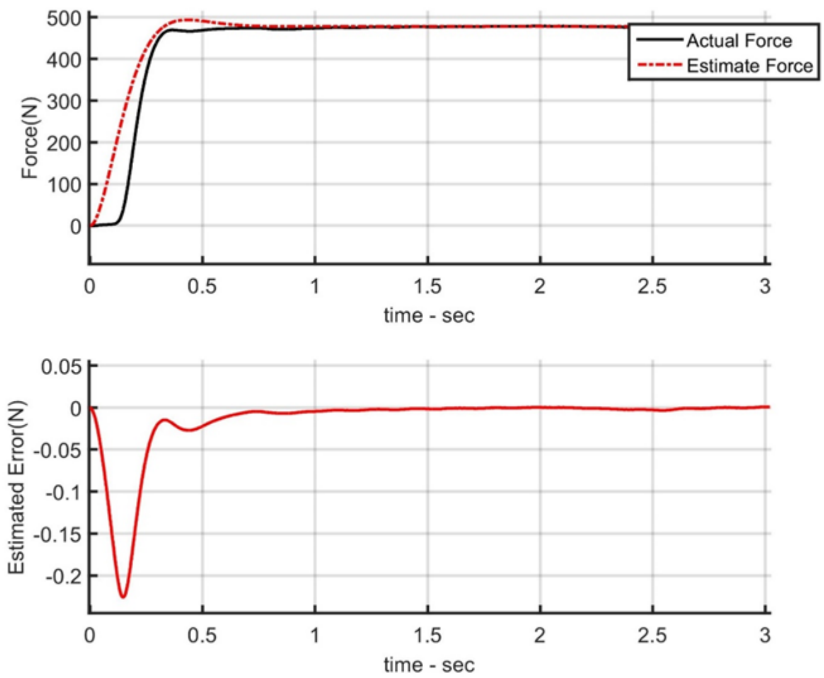


Figure 4.1: Step response of the model and the actual system.

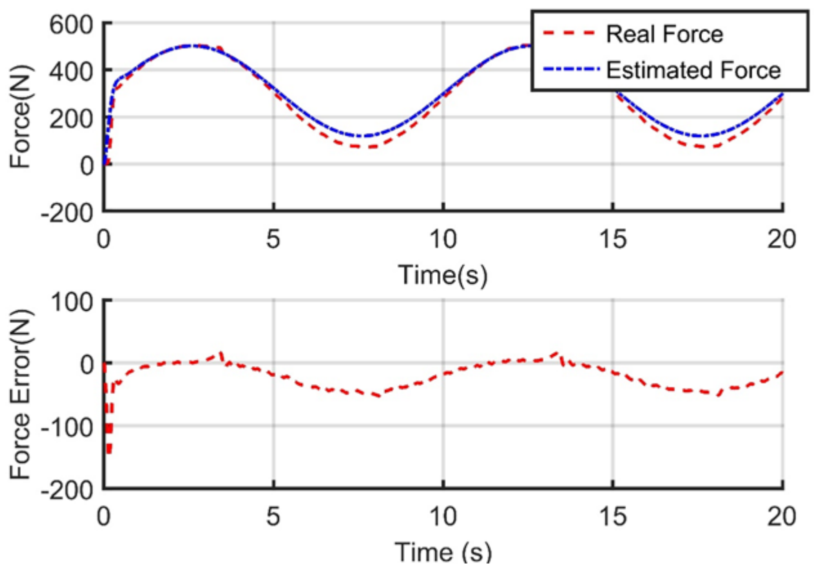


Figure 4.2: Comparison between the estimated force from model and the measurement force from muscle at frequency 0.1 Hz.

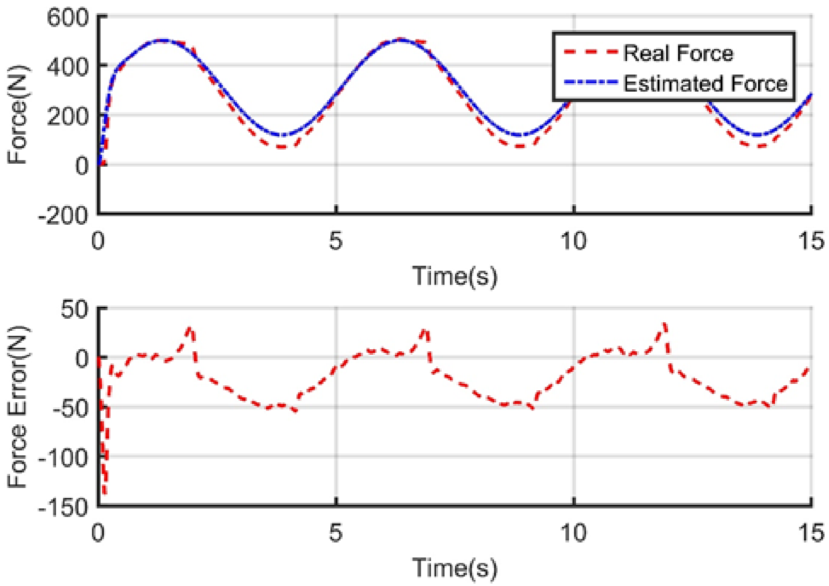


Figure 4.3: Comparison between the estimated force from model and the measurement force from muscle at frequency 0.2 Hz.

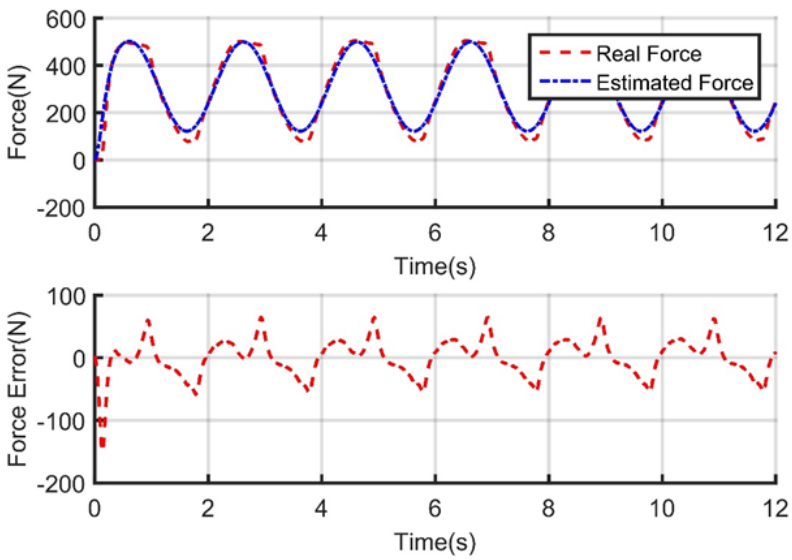


Figure 4.4: Comparison between the estimated force from the model and the measurement force from muscle at frequency 0.5 Hz.

error is bigger than the low frequencies. In the case of BWS application, the too precise control system is not necessary then the approximated simple model, in this instance, is acceptable. The key thing is that the derived model is very simple so it would be easily implemented in the real-time device which has very limited resource.

4.3 Control strategy for BWS system

4.3.1 COP tracking control scheme

Figure 4.5 shows the COP movement of the subject during walking in one gait cycle. The COP of human, at first, locates at the left leg, when this leg becomes a swing state, the COP change to the right leg (the leg is in the stance phase at this time). The subject COP will travel along of the stance phase on the right leg until the right leg becomes the swing state and at the same time the left leg becomes into the stance state the COP will again change to the left leg. The process is repeated in every gait cycle during human walking. In the case of using treadmill system, the subject is walking on it so that the position of the subject does not change. The moving of dual belts is opposite with the subject walking direction and moves the COP backward. The moving of COP is redrawn as shown in Figure on the right-side.

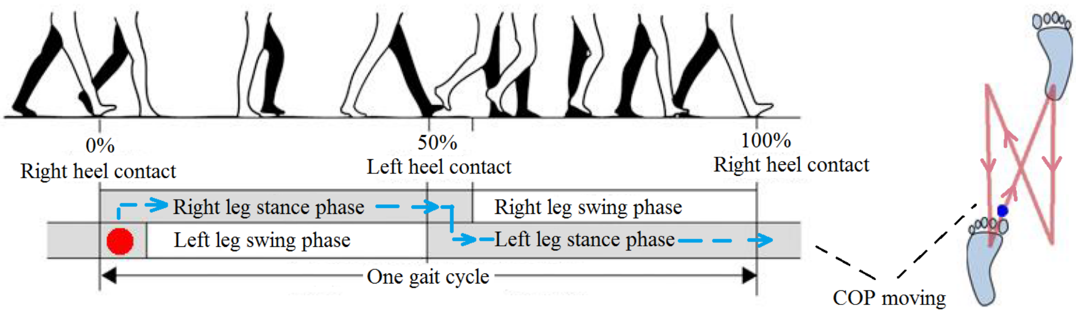


Figure 4.5: Demonstration of COP moving in force plate.

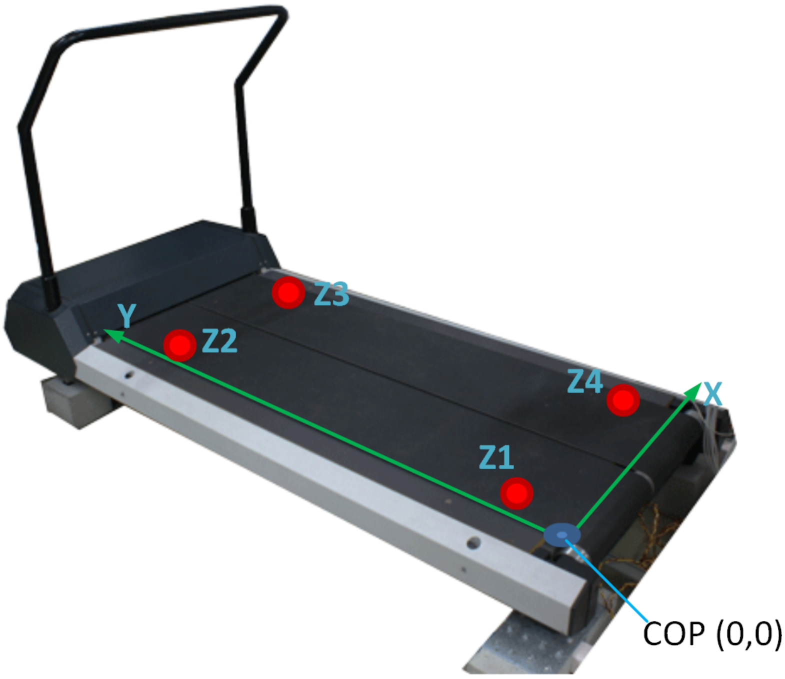


Figure 4.6: Treadmill with four force sensors integrated into four corners.

The COP tracking model was ideally based on the moving of COP in one gait cycle when subject walks on a force plate. In our system, the COP moving was identified by using a treadmill with four force sensors in four corners as shown in Figure 4.6 . Considering only the moving of COP in the x direction (COP-x), we construct the simple model that tracks the COP-x. At first, when subject stood on the Treadmill, the subject weight would be evaluated from the force sensor signals. Then, the percentage of unloading force would be input manually by increasing the pressure signals and checking the difference between the subject weight measured from the previous step and the actual value measuring from the force sensors. The obtained pressure then was set as the maximum pressure value of supplying for each PAM. Particularly, the maximal value of the input pressure for PAM mean that subject weight totally changes on its side and PAM will generate a force for unloading subject equivalent to 100% of the necessary unloading force. Then, when COP changes its position, the input pressure signals will change in both two PAMs. The input pressure value will be equal in both two PAMs when subject contributes the weight equally in both legs, in other words, the COP-x is in the center. By reducing the percentage of unloading force, the highest value of pressure will change, and the instantaneous value of pressure signal in each PAM will change following the moving of COP-x as shown in Figure 4.7 . Figure 4.8 shows the block diagram of the active tracking COP moving control model. Therefore, a simple control law for each PAM would be chosen to track COP moving following.

$$p_1 = P_m * x + b_1 \tag{4.3}$$

$$p_2 = P_m * x + b_2 \tag{4.4}$$

In the equations 4.3 and 4.3, p_1 and p_2 variables represent the instantaneous values of the pressure signal in PAMs. P_m coefficient represents the highest value of pressure input signal while x variable represents the position of the COP in the x direction. b_1 and b_2 parameters express the offset values which depend on the actual size of the treadmill.

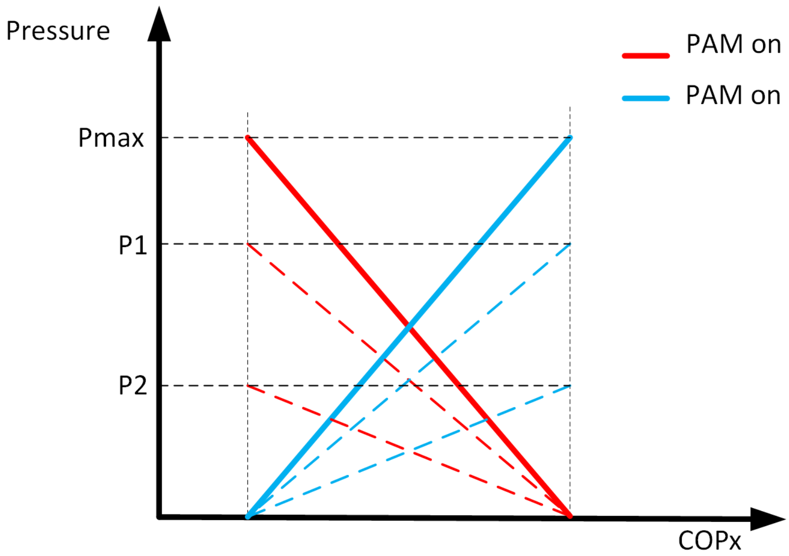


Figure 4.7: Demonstration of changing the instantaneous value of pressure in PAMs.

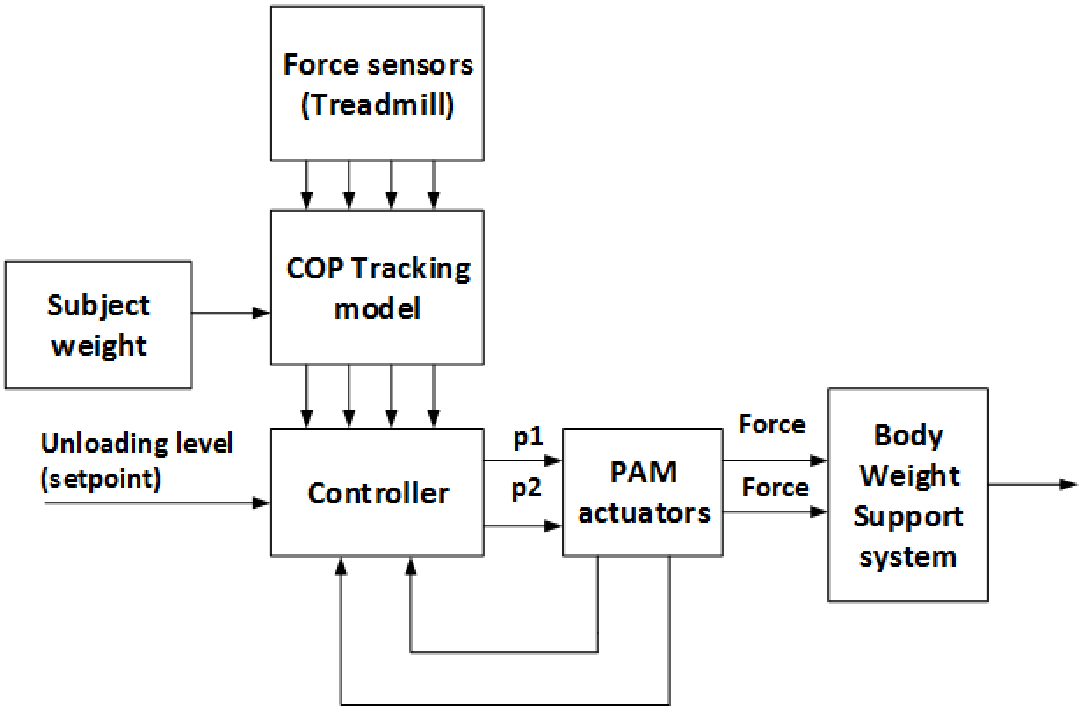


Figure 4.8: Block diagram of proposed tracking COP movement control model.

4.3.2 PID controller

In this study, conventional PID controller was used for controlling the Pneumatic muscle of BWS system. The advantage of PID controller is that it is very easy to be implemented and it is used very commonly in the mechanical system and in industrial. One of the reason author chooses the PID controller is that author tends to use the compact Rio device provide by National Instrument based on FPGA platform. The device has tiny resource and strict time clock for every loop programming. If we use other controller system strategies which would require significant resource and take much time for computing these are not meet the requirements for this device.

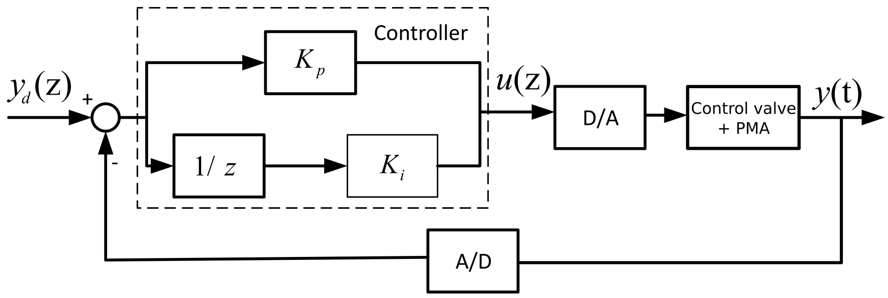


Figure 4.9: Block diagram of the PID controller.

4.4 Control system using LabVIEW Real-time and LabVIEW FPGA

4.4.1 Hardware

A key step in control design and implementation is to have a suitable device so that it can execute the algorithm smoothly and correctly. In National Instrument system, there is numerous way to implement the control algorithm, and the selected solution depend on the requirement of the application. One can use embedded device, FPGA solution or even use the general computer to implement the control program. However, the sample rate range for each solution is very different. For general computer because of the CPU has to process many tasks at a time, and the control program designed by LabVIEW is not a priority task, then control system can only run at very slow sample time. Therefore, it not a good choice for Body Weigh Support system. An embedded device is a good solution since it runs at higher frequency and reliability, the disadvantage of embedded system in comparison with a general computer is the limitation resource like CPU frequency clock or memory. For FPGA solution, the most advantage of control program implemented by using FPGA chip is that it is very high performance and reliability because FPGA in the compact Rio device is a blank chip and the program load into FPGA chip will perform like an electric circuit. FPGA program could implement at the very high sample rate. Nevertheless, the size of the program executes on FPGA is not too big and depends on the number of logic cells of FPGA chip. At this time, we have three components in case of an application for control program: general computer with huge resource and limitation of sample rate, embedded device in the balance of performance and resource and FPGA with very high performance and limitation of resources. For the BWS control system, all three components were used up for controlling and analyzing the system. The general computer was used to set the initial parameters and send every command to the real-time device. The control program was design base on FPGA platform for ensuring high performance of control system and measurement system. What data collect from measurement system was sent to the real-time device and save to the memory of the real-time instrument. The real-time device will receive every command from the user (using a general computer) and direct these command to FPGA program executing in FPGA chip.

4.4.2 Software

Figure 4.10 show the blog diagram of every composition in software program base LabVIEW for BWS system. The key methodology for software programming is that it must combine these advantages of each component from hardware control devices. For the Windows software part, the Windows machine is a general-purpose computer which has many resources such as very high CPU frequency clock, large space memory of for archiving. However, the performance of the software has significant jitter, so the program based Window platform commonly is used for data analyzing and to send every tag, message, and command to the target computer. For FPGA part, FPGA hardware could be considered as an electronic circuit then it could operate at the very high frequency and very low jitter for a single cycle time. Then, the FPGA program which included control program and acquisition program is embedded inside the FPGA chip to get the highest performance and reliability. Particularly, the FPGA in compact Rio is a white chip. Since the FPGA software is compiled, deployed and executed, one processing inside FPGA chip is rewired on hardware to implement that functionality that defined in software. For the real-time target, it has a balance between resource and performance. However, the recommendation sampling rate for an application using the real-time device is lower than 500Hz.Therefore, it does not meet the requirements for control and measurement system in case of BWS system. Hence, real-time target device would take responsibility to connect between the user computer and FPGA part inside the compact device and to log data which are collected by FPGA program. Network publish shared variables are used for communication between Windows program and Target program because we want to send only the latest values like updating parameters or messages. In communication between Target program and FPGA program, FIFO memories are used because we want to send all collected data from a sensor for analyzing progress later. Figure 4.11 shows the LabVIEW project and User Interface of Window program which is used for control the BWS system.

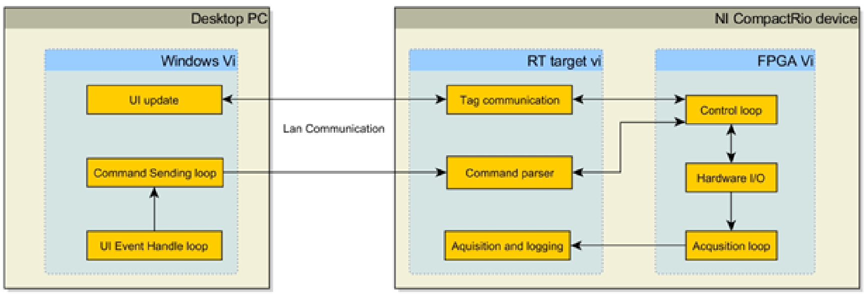


Figure 4.10: The main blocks of software program based LabVIEW using for controlling the BWS system.

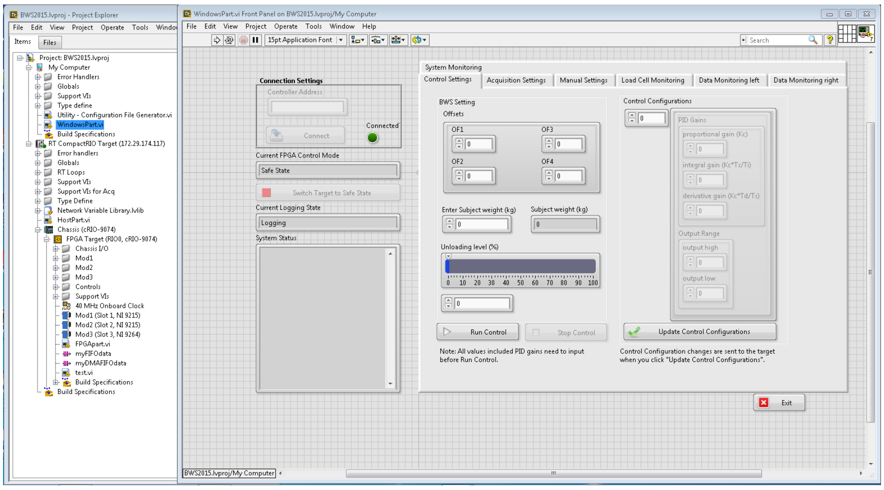


Figure 4.11: LabVIEW project and User Interface of the software program for controlling BWS system.

4.5 Conclusion

This chapter proposes the control system scheme and controller for the novel Treadmill BWS system. In this section, the COP tracking model of the control scheme and the strategy of the unloading weight support were introduced. The application of the COP tracking model is totally a new strategy for generating the unloading force applying for rehabilitation. All the kinematics analysis and the proposal derivation of the COP tracking model that generates the input reference for the actuators were described in details. Furthermore, the controller design and a proposal method for control implementation base on LabVIEW real-time and LabVIEW FPGA were also described.

Chapter 5

Result and Discussion

The fifth chapter consists of the results and discussion of this research project. The first sub-topic would describe the experiment design for system validation. After that, several sub-topics delves into the assessments evaluation results and analyzing. In this chapter, the validation experiments were designed including BWS system and Counter Weight system for comparison. The motion capture system with six cameras also was used for recording twelve reflexed markers to calculate COM for comparing between the new system and the old one. Moreover, in the experiment assessment for the new BWS system, unloading forces, reaction forces, COP also were recorded at several weight unloading levels for both BWS system and Counter Weight system so that we could have the whole picture of the advantages and disadvantages of the new system.

5.1 Experiment protocol

Nine healthy subjects (all subjects are male with age = 24.2 ± 3.2 (mean $\pm$ SD), height = 172.1 ± 6.1, and weight = 61.5 ± 6.51) with no prehistoric disability took part in this experiment to validate the system. We conduct these experiments to determine the intrinsic differences between the Counter Weight system and the new proposed BWS systems. We are interested in checking if the new proposed BWS systems can reproduce the behavior of a normal walk. For each subject, we did experiments with different body weight unloading forces, mainly 20%, 30%, 50%, 70%, and normal walking for both BWS system and Counter Weight system. In the first ten seconds of the experiment, we measure the subject's weight by asking to him to maintain a right posture (same weight on both legs) and on the middle of the treadmill. Then the subject started to walk on the treadmill at a speed of 1 km/h for 60 seconds. The procedure is iterated for every experiment as well as for all unloading force levels. The data collected from four force embedded sensor was used to calculate the reaction force and the COP parameters. The motion capture system with six cameras also was used to collect data from 12 markers positioned at the ends of the bone of lower limbs, pelvis, and shoulder of the subject. The data collected from the motion capture system were used to calculate the COM moving. COM is considered very carefully for this research project because for using a BWS system or Counter Weight System the COM pattern is much affected. The time series data that collected from the sensors is difficult for analyzing the difference among the data of cases normal walking, BWS system, and Counter Weight system. A Paired t-test was used for analyzing the difference of the mean of data for a representative subject. In order to analyze the difference of the gait for all nine subjects in cases of normal walking, BWS system, and Counter Weigh system, the gait parameters such as COP amplitude in mediolateral (step width), COM amplitude (in both mediolateral and vertical) were quantified. A standardized procedure was applied to all parameters of all nine subjects. The standardized data then was used to analyze the difference among the BWS system, Counter Weight system, and the Normal walking case using one-way ANOVA method. The Tukey HSD test was used to investigate the difference between each couple cases. The procedure of analyzing was conducted based on the R statistical language using R Studio (a free open source software).

Two MATLAB routines have been developed to analyze the gait parameters such as vertical ground reaction force (Fz), COP trajectory, and COM movement. The Figure 5.1 shows the detail of the first routine to calculate the Fz and COP trajectory in one gait cycle. The data collected from the sensors and the motion capture system are commonly in waveform and combine many gait cycle together. The key point to separate the COP data to every individual gait cycles is to

Figure 5.1: The procedure to calculate vertical ground reaction force (Fz) and COP movement.

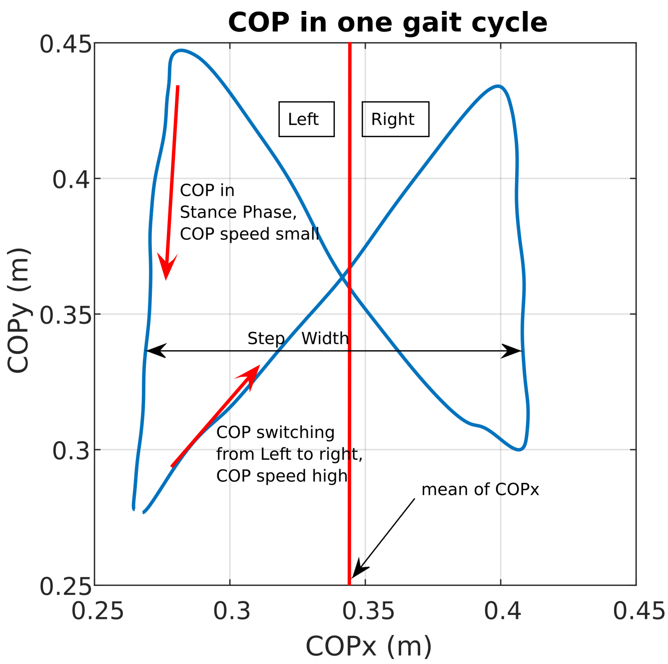


Figure 5.2: COP in one gait cycle and the definition of step width.

determine the event that appears during locomotion. In the case of COP and Fz calculation, the event that starts a gait cycle (heel strike is considered). The Figure 5.2 shows the analyzing of COP moving when right heel strike during gait. When the right leg heel is striking, the COP switches from the left to the right side. We notice that the COP in x coordination (COPx) at right heel strike is smaller than the average value COPx. The event that the COP is switching from the left to the right side (right heel strike) could be identified by the velocity of the COP. Notice that the COP during left leg stance phase is very small; when COP start to switch from the left to the right, the velocity of COP is rising very high. The moment, such that the COP velocity is rising very high and the COPx smaller than the mean value of the COPx, is defined as the right heel strike (and so called as Heel-strike detection) and is used to separate the data to every individual gait cycles. The step width, which is used as a parameter for quantifying the COP parameter, is defined as the distance between two edges (left and right) of the COP trajectory. The Figure5.3 shows the procedure of the second MATLAB routine to calculate the COM parameters in one gait cycle. In the case of calculation of COM in one gait cycle, the event to detect the

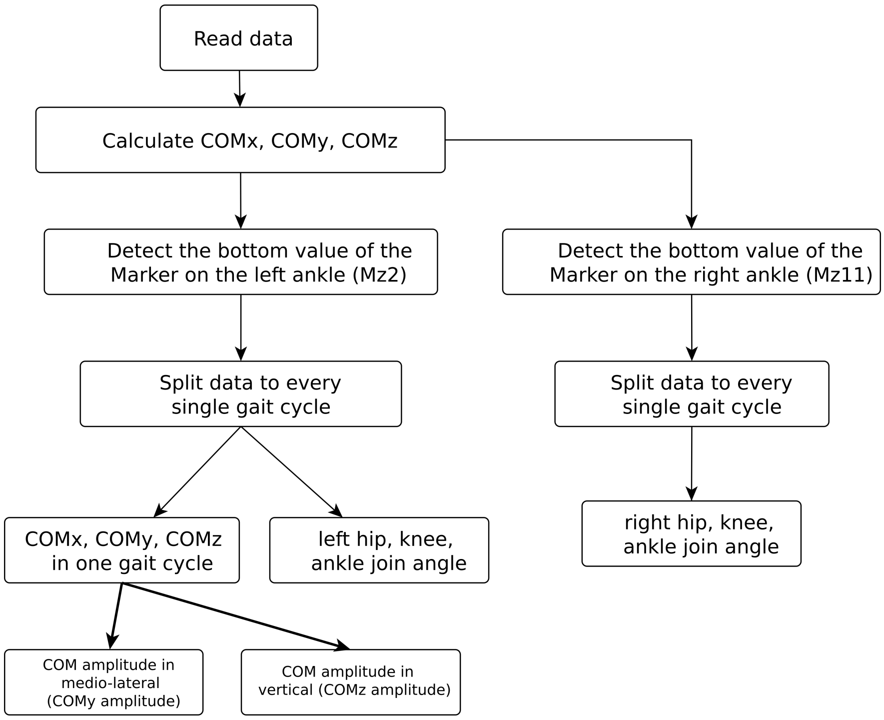


Figure 5.3: The method to partition the COM data and to quantify the COM parameters.

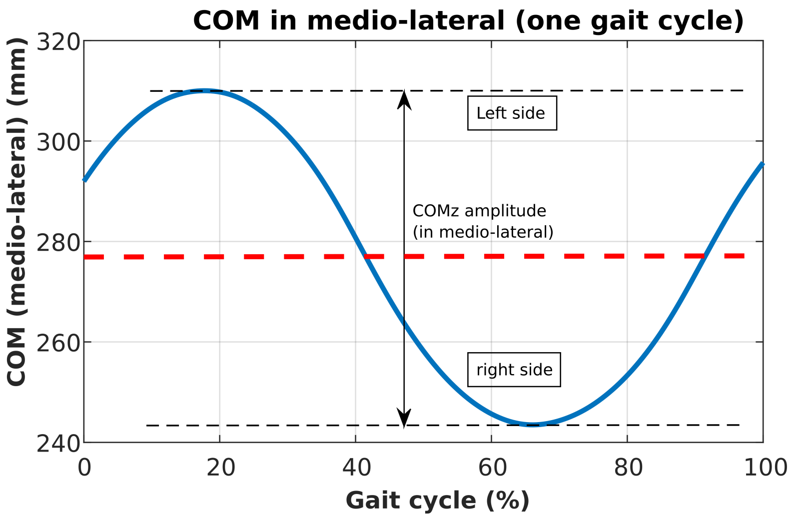


Figure 5.4: COM in mediolateral direction and the definition of the COMy amplitude in mediolateral.

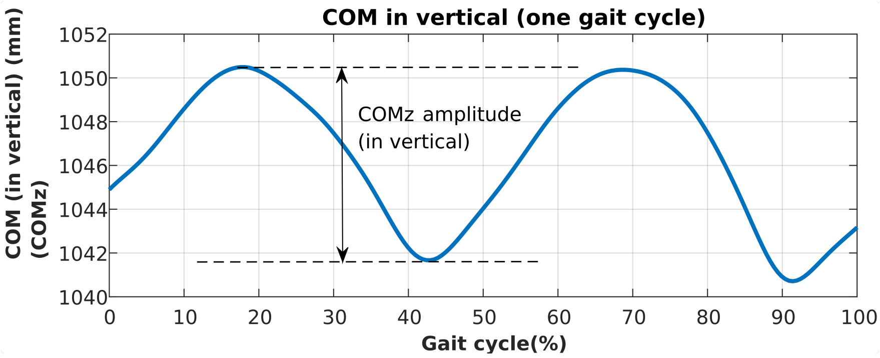


Figure 5.5: The COM in the vertical direction and the definition of the COMz amplitude in vertical.

starting point of a new gait cycle is defined as the marker on the ankle joint goes down. In that case, the event that the coordination in vertical of the marker on the ankle joint goes to a minimum value is chosen as the detection point of a new gait cycle. From the detection point, the data of COM is partitioned to every individual gait cycles. The COM amplitude parameters in mediolateral and vertical (as shown in the Figure 5.4 and Figure 5.5) are quantified to represent the effect of the BWS system and Counter Weight system. The data of COP and COM amplitude parameters then were standardized for all nine subjects. The standardized procedure for the experimental data of each subject is conducted by using the formula below:

$$Standardized\ value = \frac{parameter\ value(Normal, BWS, and\ CWScase)}{Mean\ value(Normal\ case)} \tag{5.1}$$

The standardized data then was used to analyze the difference among the BWS system, Counter Weight system, and the Normal walking case using one-way ANOVA method. The Tukey HSD test was used to investigate the difference between each couple cases. The procedure of analyzing was conducted based on the R statistical language using R Studio (a free open source software).

5.2 Reaction force and unloading force

Figure 5.6 the reaction forces of the representative subject recorded when the subject is walking on the treadmill under unloading force at 30%, 50%, and 70% weight unloading. The blue line represented for the reaction force when subject walking using the BWS system, the green line represented the reaction force when subject walking using the counterweight system and finally the black line is for normal walking (mean that 0% weight unloading and the subject did not wear the harness as well). The reaction force in normal walking is for comparing among weight support systems and locomotion posture at which subject felt most comfortable. We could see that both BWS system and counter weight system modified the subject's reaction force and made reaction force pattern deformation. When the weight support level increased, the deformation of the reaction force patterns was also increased. This change of the reaction force pattern is because at the high levels of weight support subject would be difficult to touch the whole sole (include both heel contact and toe off) on the treadmill. Moreover, the higher weight support also affects stronger to every gait parameter so it would be difficult to say reaction force would have scaled pattern from normal walking. Intuitively, we may see the reaction force at low and medium weight support for the BWS system look like better reaction force pattern than the Counter Weight system. However, the differences in the reaction force patterns were not clear between two weight support systems.

The reaction force patterns in Figure 5.6 represents for many gait cycle, and it would be difficult to comparing the effect of the weight support systems. Therefore, to observe the reaction force easier, the reaction force patterns were split into separated single gait parts, and finally, the reaction force patterns were averaged for all single parts. Because of the number of points for each reaction force pattern of the weight support system were different then the horizontal axes represent the percentage of gait cycle for easily comparing. The method was also applied to the following resources. The Figure 5.7 and Figure 5.8 represented for the reaction force in one gait cycle at three levels of weight support 30%, 50% and 70%. In Figure 5.8, the reaction force in case of normal walking was scaled

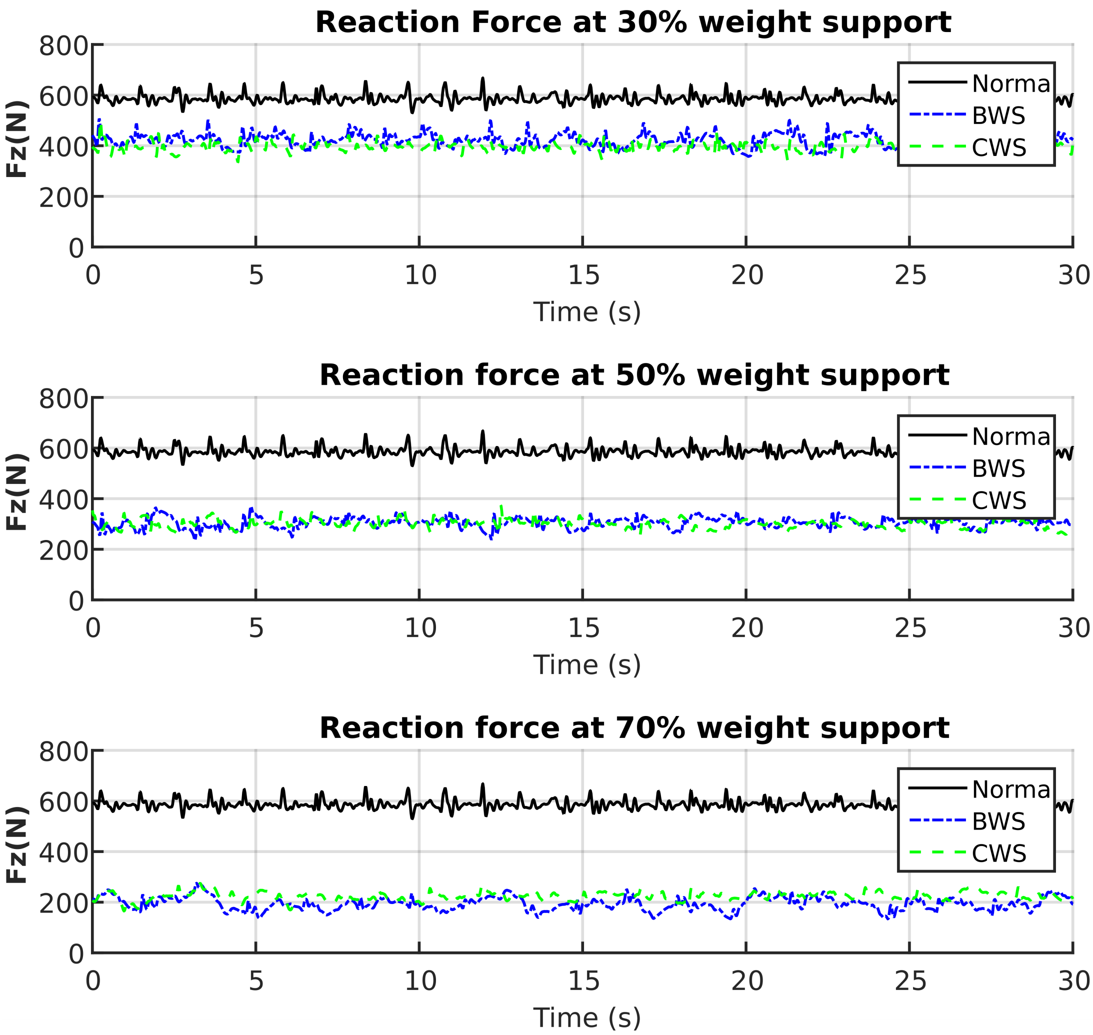


Figure 5.6: Reaction Force When applying BWS system. The black line represents for the normal walking, the blue line represents for BWS system, and the green line represents for Counter Weight system. From the top to the bottom, the weight support level is 30%, 50%, and 70%, respectively.

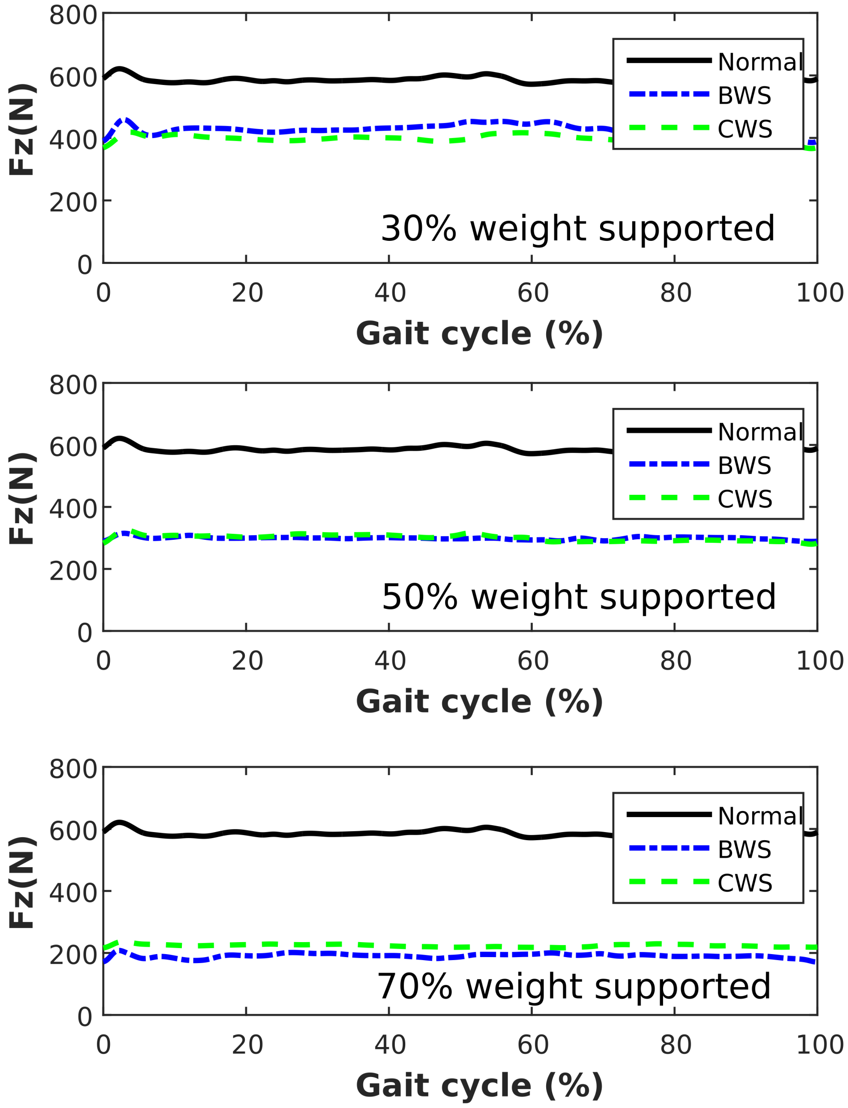


Figure 5.7: Reaction Force in one Gait Cycle the black line represents for the normal walking, the blue line represents for BWS system, and the green line represents for Counter Weight system. From the top to the bottom, the weight support level is 30%, 50%, and 70%, respectively.

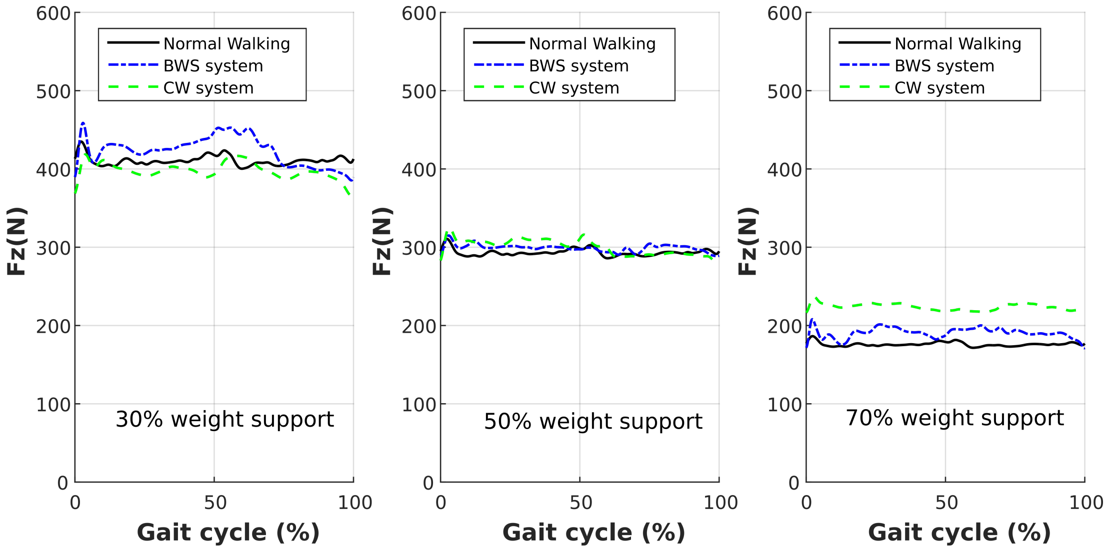


Figure 5.8: Reaction Force When applying Weight systems. Backline Reaction Force at 0% is scaled for comparison, Blue Reaction force using the Active system, Green Reaction force using Counter Weight system. From the left to the right, the weight support level is 30%, 50%, and 70%, respectively.

0.3, 0.5 and 0.7 times so that we could see how modification of the reaction force pattern by weight support systems in comparing with the normal walking. In almost case, we may see reaction force shapes for both Body Weight system and counter Weight system were matching the normal walking reaction force pattern. In the Figure 5.8, the reaction force in case of using Body Weight system was higher matching to the normal walking than the counter weight system.

In the table 5.1, the mean difference expresses the difference of reaction force pattern of the representative subject in the case of applying weight support system with the reaction force pattern in normal walking (scale to the percentage of weight support). We could see that every reaction force pattern differs from reaction force in normal walking, what we could consider here is how much of the difference. In the case of the low level of weight support every reaction force difference is about 10(N) to 15(N), the difference by BWS is a little bit higher than the case of CW system. At a high level of weight support the differences of reaction force increase for both cases of weight support system. However, the difference by applying the Counter Weight system is significant. The negative

Table 5.1: Comparison Reaction force between BWS system, counter Weight system and Normal Walking of the representative subject by Paired t-test.

Method	Weight Support	Sample Size	Mean difference (N)	SD	SE mean	95% CI down	95% CI up
BWS	30%	8800	11.242	27.864	0.297	10.66	11.825
CWS	30%	8800	-13.279	21.775	0.232	-13.734	-12.824
BWS	50%	6450	15.853	20.144	0.251	15.362	16.345
CWS	50%	6450	9.636	18.013	0.224	9.196	10.076
BWS	70%	6752	19.014	25.382	0.309	18.408	19.619
CWS	70%	6752	50.963	16.522	0.201	50.569	51.357

behavior of CW system is due to the "pendulum effect" increase strongly at the high level of weight support.

The Figure 5.9 expresses more clearly the mean difference of Reaction force patterns of the representative subject during walking under in case of comparison to reaction force in normal walking. In this Figure, the vertical axis represents the difference in the reaction force patterns; the more closed to 0 is the more agreement between two patterns. The negative value expresses the lower amplitude of the pattern to the reference pattern; the positive value expresses the higher amplitude. The horizontal axis expresses the weight support level. We could see that in the case of using BWS system, the difference is increased steadily since the weight support level increases. In the case of using Counter Weight system, the difference increases extremely when the weight support level increases. These results also confirm that the "pendulum effect" strongly effects to gait parameters at the high-level weight supports.

In Figure 5.10 and Figure 5.11, the adaptation of the unloading force was expressed for the left leg side. For the reference input signal (named Refinput), that one is the desired unloading force on the left-side which depends on the position of COP since it is on the left-side. When the COP is changing to the left side, the unloading force also is rising to the top. This rising is very fast because the COP switches very quickly. At the stance phase, the COP travels along the

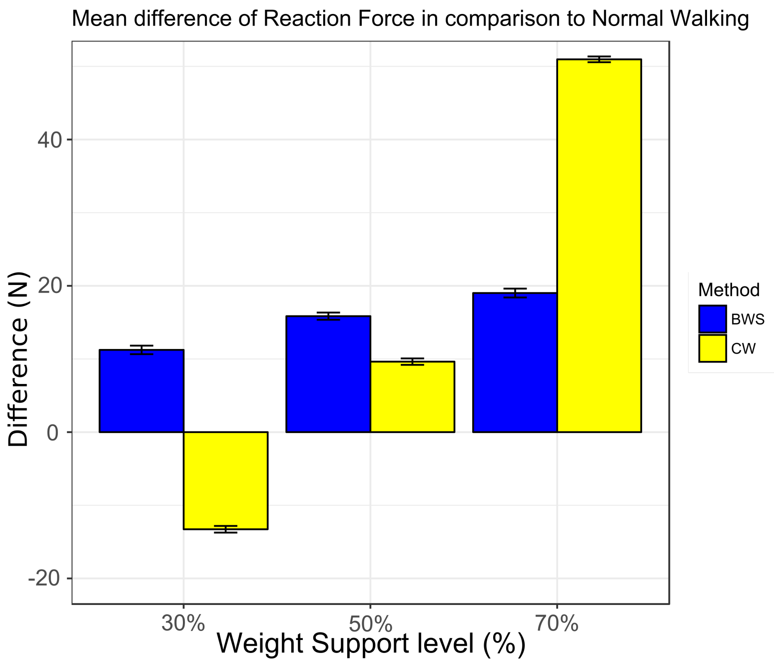


Figure 5.9: The demonstration of the difference of reaction force pattern when using the weight support systems to reaction force pattern in normal walking of the representative subject. Difference = Fz by weight support system - Fz by normal walking.

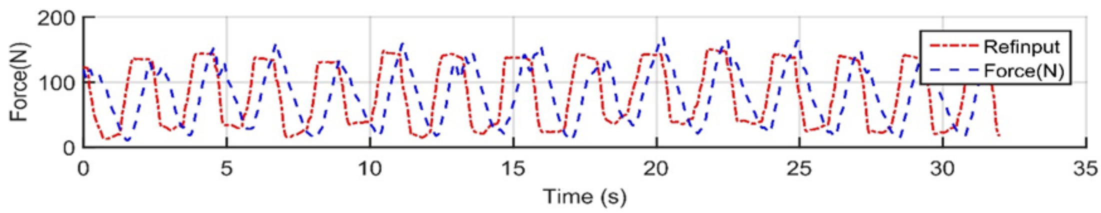


Figure 5.10: The adaptation of the unloading force for the BWS system on the left side. The red line represents for desired unloading force (named Refinput), the blue line represents the actual force.

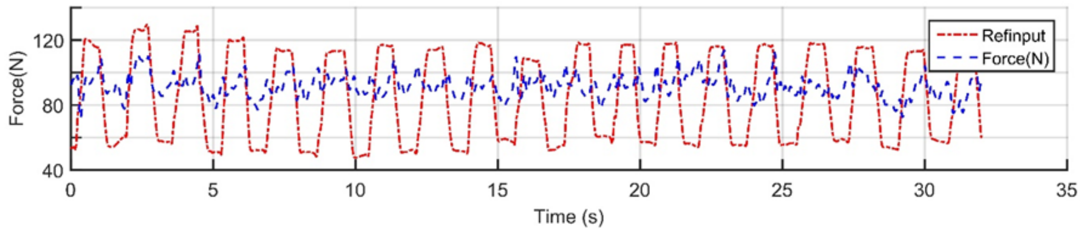


Figure 5.11: The adaptation of the unloading force for the Counter Weight system on the left side. The red line represents for desired unloading force (named Refinput), the blue line represents the actual force.

sole from the heel to the toe and this moving takes time then the unloading force keeps as a constant there. When the COP switch to the right and locomotion posture changes to swing phase, the unloading force on the left side goes quickly down. The same calculation is applied for the desired unloading force in case of the counter weight system. The blue line represents the actual unloading force recorded by the left load cell mounted in the left hardness bell. We may see that the real force tries to follow the desired unloading force in Figure 5.10. For the lower Figure, the real force does not follow the desired unloading force. Instead, it tries to be a constant.

5.3 COP trajectories

Figure 5.12 shows COP trajectories with 30%, 50%, and 70% of the subject weight for the representative subject. Each experiment was conducted using both two kinds of BWS system including Counter Weight system and BWS system using the COP tracking model. We also recorded a normal walk (with 0% of unloading force) to give a graphical comparison of how the real COP path without the BWS system must be. In Figure 5.12, the black line represents the COP trajectory without unloading force while blue and green lines represent the COP trajectories for counter weight and BWS system with the active model, respectively.

From Figure 5.12 we can see that the COP trajectory of the subject using the active BWS system is more similar in normal walking than the other. Especially in the lower level of unloading force, the active BWS system shows its best behavior comparing with the counter weight system. The responses from subject revealed that subject's feeling is more comfortable using the active BWS support system than the counter weight system. On the other hand, results show significant differences between the COP trajectories of the subject in normal walking and when using counterweight system (consider in the x direction). When considering COP paths of BWS systems and the counterweight system in the higher level of the unloading force, the difference of the COP paths is reduced. It is because at the high degree of weight support subject would difficult to touch the whole sole especially the heel contact so the COP paths are deformation and therefore we may difficult to see the difference between them.

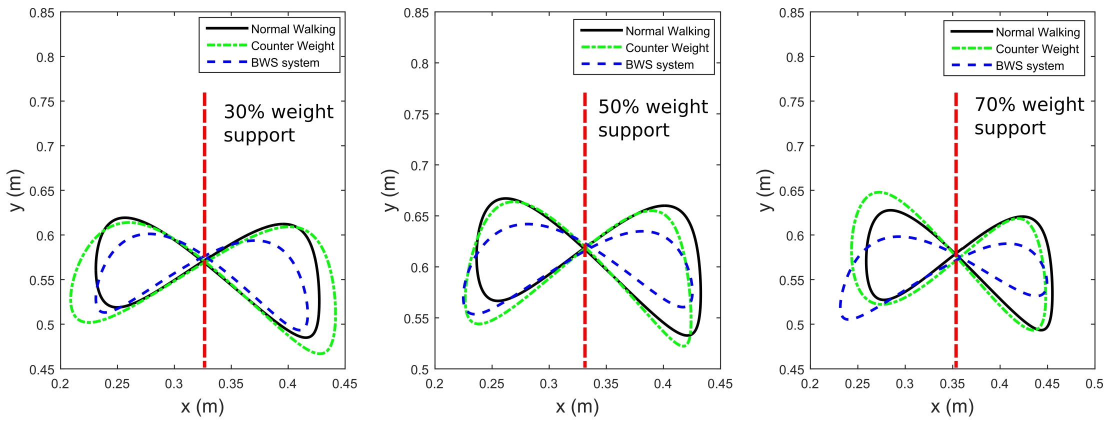


Figure 5.12: COP trajectories of unloading subject weight of the representative subject. The black line represents for the COP at normal Walking, the Blue line represents for Counter Weight system, and the Green line represents for BWS system.

In order to quantify the COP in the lateral direction, the step width is considered, the data of step width of all nine subjects which is standardized to the normal walking case was used to investigate the effect BWS system and the counter weight system using ANOVA. The Table 5.2 represents the data of the standardized step width data in cases of the normal walking, BWS system and the Counter Weight system in which the unloading force equal to 30%, 50%, and 70% subject weight respectively. The Figure 5.13, Figure 5.14, and Figure 5.15 depicts the difference of the COP amplitude in the frontal direction in which the

Table 5.2: Comparison of the step width (of all nine subjects) between BWS system, counter Weight system, and Normal Walking by ANOVA. NormalStd, BwsStd, and CwsStd are data of step width standardized in cases of Normal walking, BWS system and Counter Weight system respectively.

Weight Support	Method	Sample Size	Mean	SD	95% CI
30%	NormalStd	139	1.0000	0.0753	(0.9756, 1.0244)
	BwsStd	139	0.7971	0.1726	(0.7727, 0.8215)
	CwsStd	139	0.8012	0.1698	(0.7768, 0.8256)
50%	NormalStd	132	1.0000	0.1131	(0.9749, 1.0251)
	BwsStd	132	0.7806	0.1510	(0.7555, 0.8057)
	CwsStd	132	0.7567	0.1697	(0.7316, 0.7817)
70%	NormalStd	138	1.0000	0.0695	(0.9644, 1.0356)
	BwsStd	138	0.9334	0.2223	(0.8979, 0.9690)
	CwsStd	138	0.9109	0.2852	(0.8753, 0.9464)

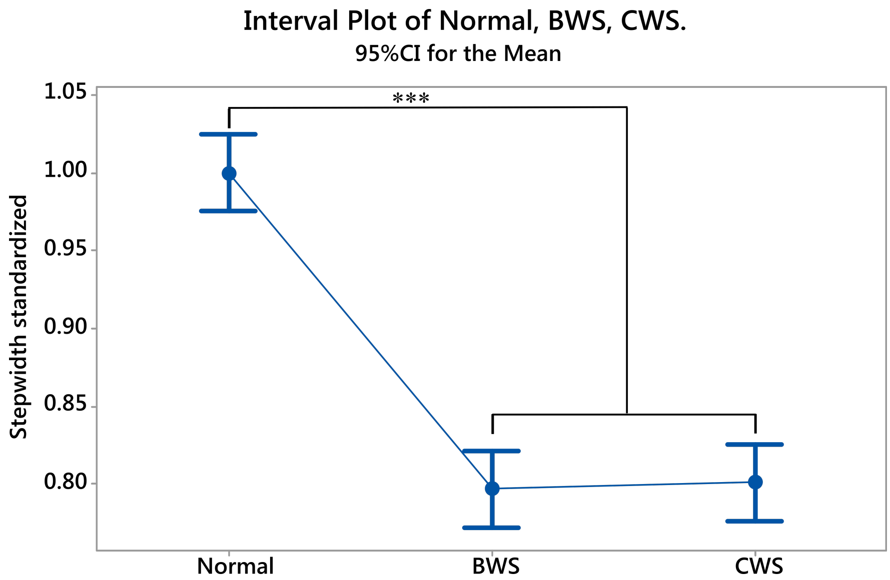


Figure 5.13: The demonstration of the difference of COP amplitudes in the frontal direction (step width) of all nine subjects when using the weight support systems at 30% weight support, *** represented the significant value $p < 0.001$.

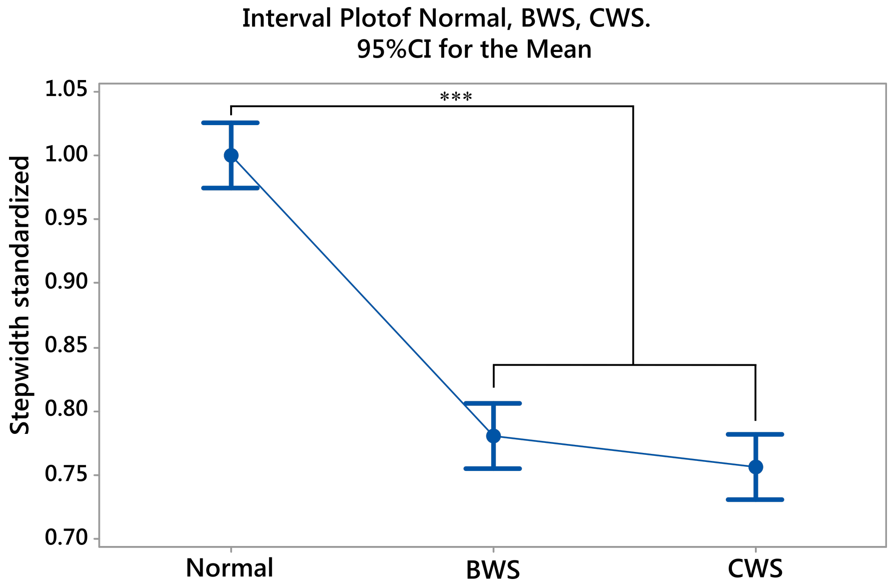


Figure 5.14: The demonstration of the difference of COP amplitudes in the frontal direction (step width) of all nine subjects when using the weight support systems at 50% weight support, *** represented the significant value p < 0.001.

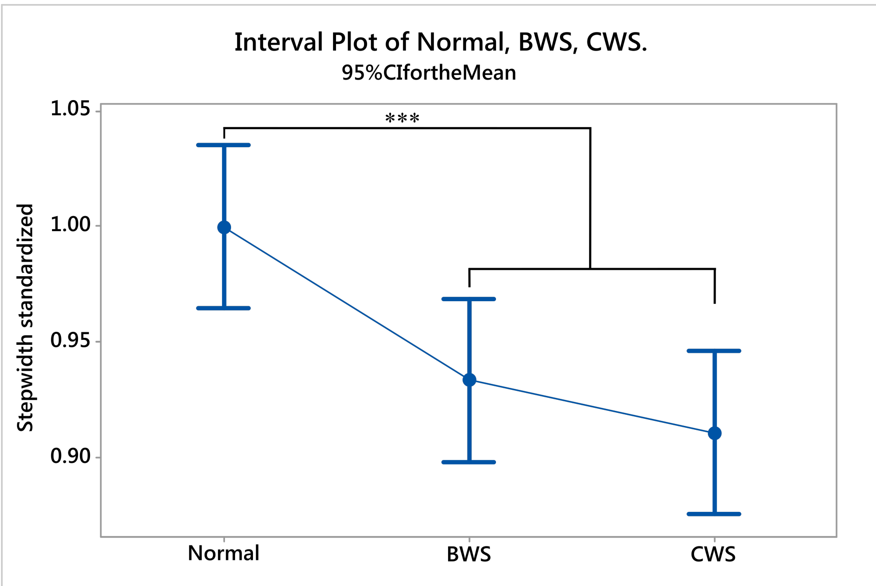
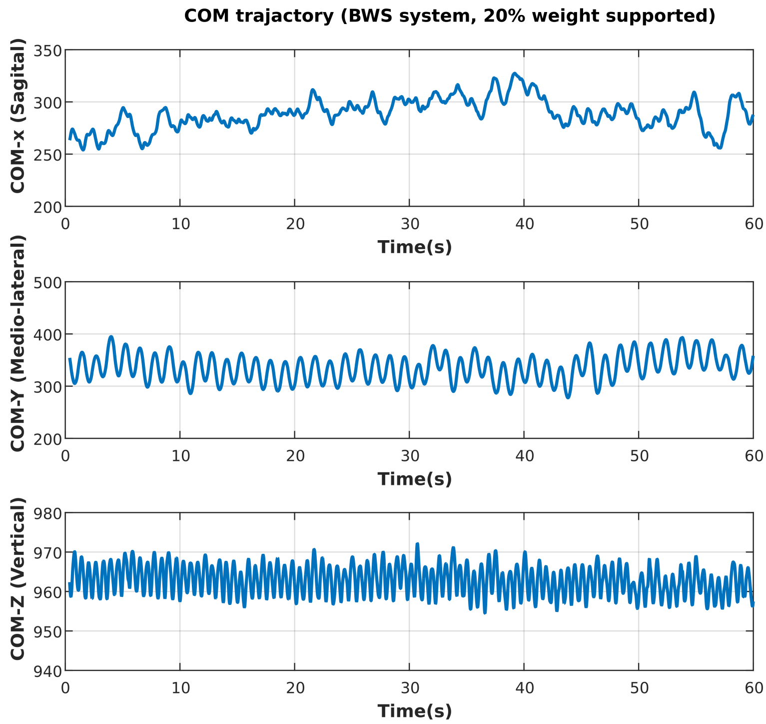


Figure 5.15: The demonstration of the difference of COP amplitudes in the frontal direction (step width) of all nine subjects when using the weight support systems at 70% weight support, *** represented the significant value p < 0.001.

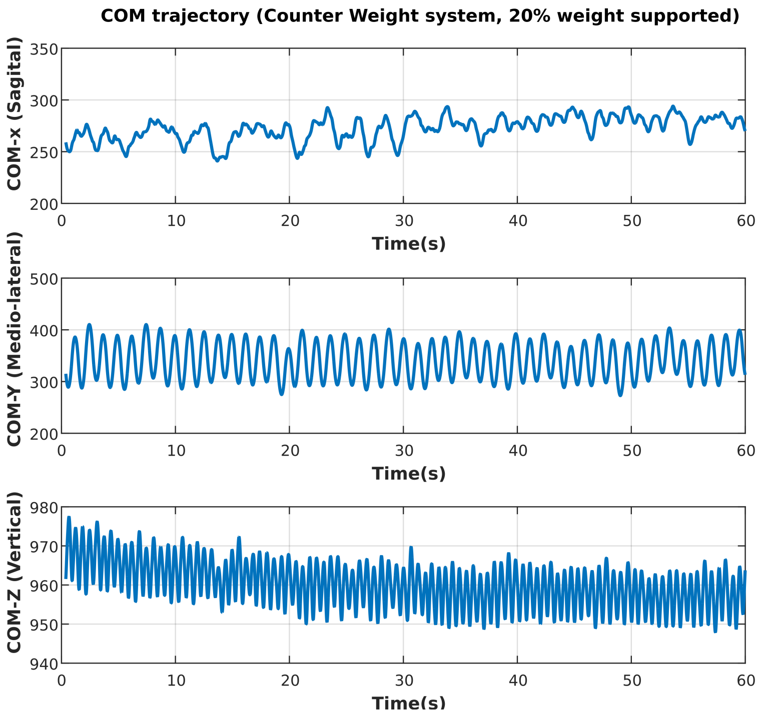
unloading force equal to 30%, 50%, and 70% subject weight respectively. These Figures show the significant difference of step width of the normal walking case to the step width during applying the BWS system and the Counter Weight system ($p < 0.001$). We could see that at the mean value of the COP amplitudes in frontal in case of BWS system are almost closer to the normal walking and higher than the Counter Weight system case. However, the significant of COP amplitude in the frontal direction between the BWS system and the Counter Weight system was not found.

5.4 COM movements

Figure 5.16, 5.17, and 5.18 show the COM trajectories in three dimensions: sagittal (COMx), mediolateral (COMy), and vertical (COMz) for both BWS system and Counter Weight system at three level of unloading force 20%, 30%, and 50% of the representative subject. The data were recorded by using the motion capture system with twelve markers at the lower limbs, pelvis, and trunk. The COMx represents the moving of COM to the sagittal axis; this means that the subject trunk trends to moving forward and backward (because of a subject walking on the treadmill). The COMy represents for the moving of the COM to the frontal axis, what we can figure out that is the swing of the COM from left to right and vice versa (mediolateral). Finally, COMz represents for the oscillation the COM to the vertical direction. Comparing three movements of COM we may see intuitively that the moving of COMz and COMy were most affected by the weight support system, and we would consider the changing of these. We could see that for all weight support system the COM in vertical direction shows quite similar at the lower level of weight support. At the high degree of weight support, the COM in the case of BWS system shows its good behavior and keeps the same shape look like the low level of body weight support. However, in the case of Counter Weight system, it looks like an oscillation was added to the COM trajectories, and COM in the vertical direction was too different in comparing the lower weight support levels and also the BWS system.

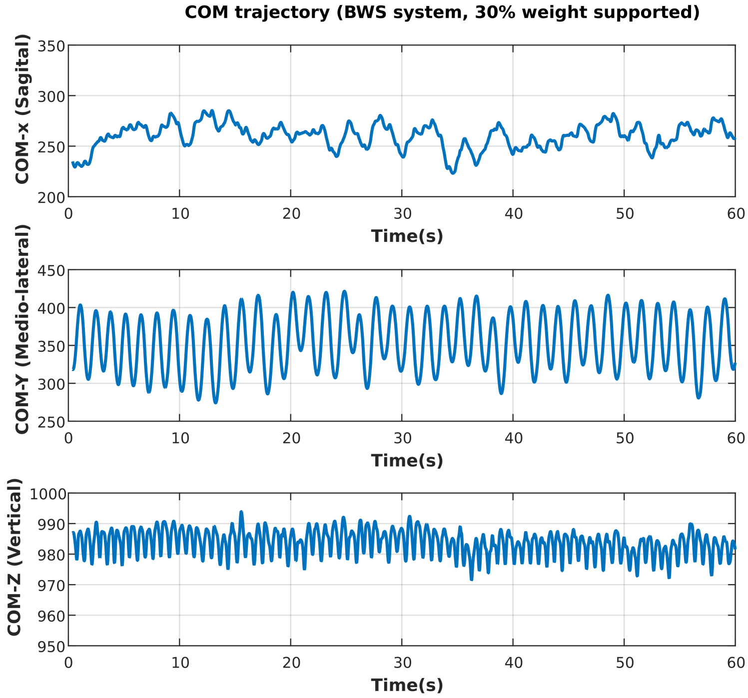


(a) COM trajectory at 20% using BWS system.

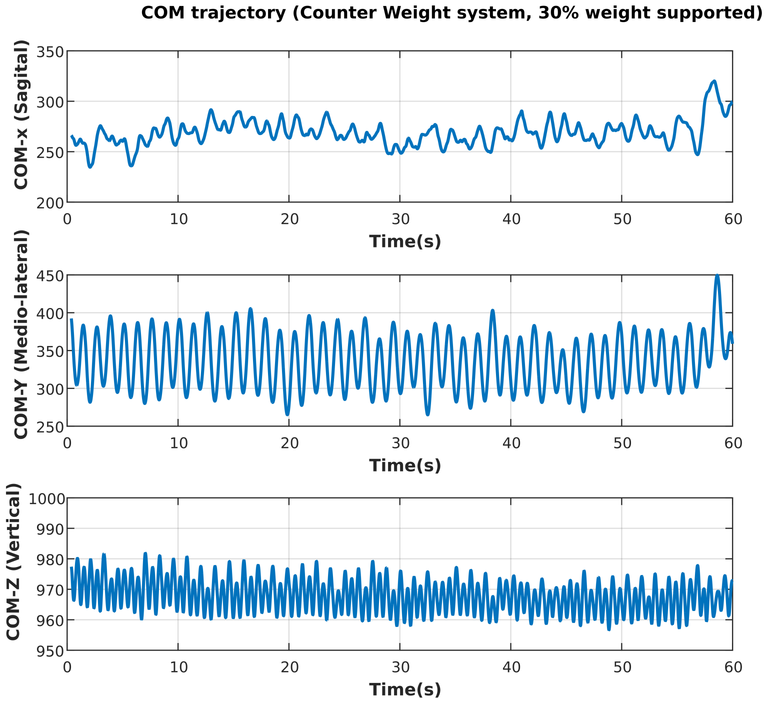


(b) COM trajectory at 20% using Counter Weight system.

Figure 5.16: The COM trajectories in three dimensions: sagittal (COMx), medio-lateral (COMy), and vertical (COMz) for both BWS system and Counter Weight system at the unloading force 20% of subject weight of the representative subject.

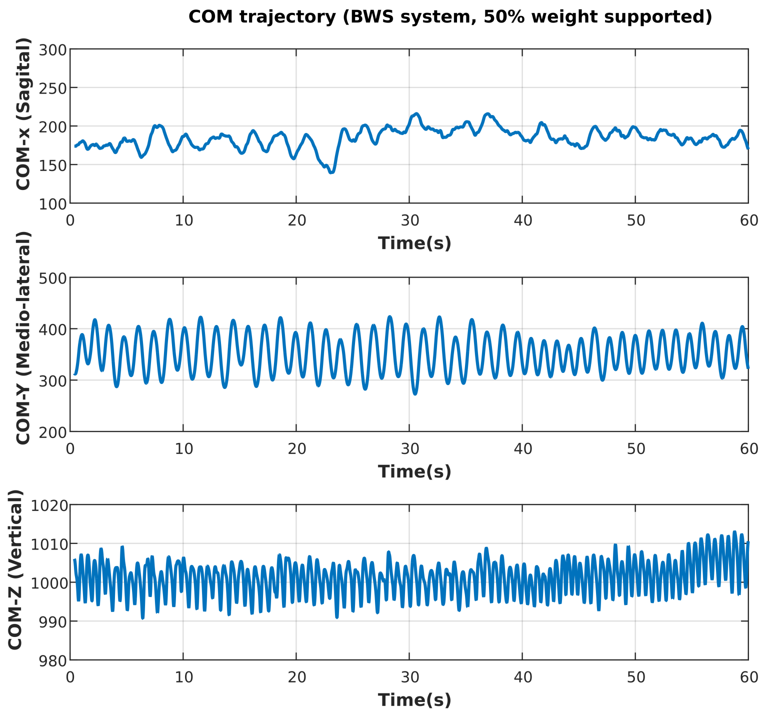


(a) COM trajectory at 30% using BWS system.

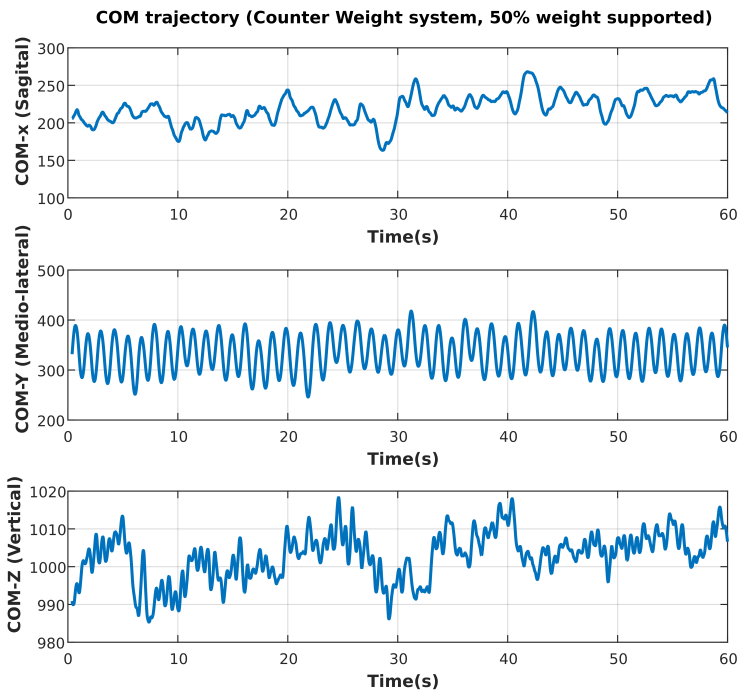


(b) COM trajectory at 30% using Counter Weight system.

Figure 5.17: The COM trajectories in three dimensions: sagittal (COMx), medio-lateral (COMy), and vertical (COMz) for both BWS system and Counter Weight system at the unloading force 30% of subject weight of the representative subject.



(a) COM trajectory at 50% using BWS system.



(b) COM trajectory at 50% using Counter Weight system.

Figure 5.18: The COM trajectories in three dimensions: sagittal (COMx), medio-lateral (COMy), and vertical (COMz) for both BWS system and Counter Weight system at the unloading force 50% of subject weight of the representative subject.

The Figure 5.19, 5.20, and 5.21 show more clearly the effects of the weight support systems since the COM trajectories were represented in one gait cycle. The method to calculate the COM was also similar to the method that calculates the reaction force and COP for one gait cycle. This means that the COM patterns were split to every single gait cycle, after that the COM trajectories were averaged for all parts. The COM trajectories also were represented to the percentage of gait cycle for easier comparing. To see more clearly, the author also added the standard deviation for all COM trajectories. In this case, we also should put more attention to the COM trajectories in the vertical direction because the weight support systems give much effect to this pattern. We could see that in general when the unloading force increases the COM in vertical direction also increases. However, the COMs in vertical direction keep the similar shape, amplitude and the variation of trajectory. At the high level of weight support, the Body Weight support system still kept the similar effect to the COM trajectory in the vertical direction. In contrast, the COM trajectory in case of the high level of unloading force using Counter Weight system was much deformation and variance, the range from negative standard deviation to positive standard deviation was bigger than the case of using BWS system.

In comparison of COM trajectory in the frontal direction of the representative subject, the variance of the difference of COM trajectories is quite similar to every case of weight support level as shown in Table 5.3. The mean difference for every case is negative this represents the amplitude of COM trajectories since applying the weight support systems is smaller than in the case of normal walking. Alternatively, in other words, we could say that the amplitude of COM trajectories in frontal direction when applying weight support system will reduce. The difference of COM in the case of using Counter Weight system increases (from 16(mm) to 26(mm)) when the weight support level increases (from 20% to 50%). However, in the case of using BWS system the difference in the most case of weight support level is quite small and at the high level of weight support level, the mean difference is much smaller than Counter Weight system. This one represents that COM by using BWS system is much similar to normal walking than Counter Weight system.

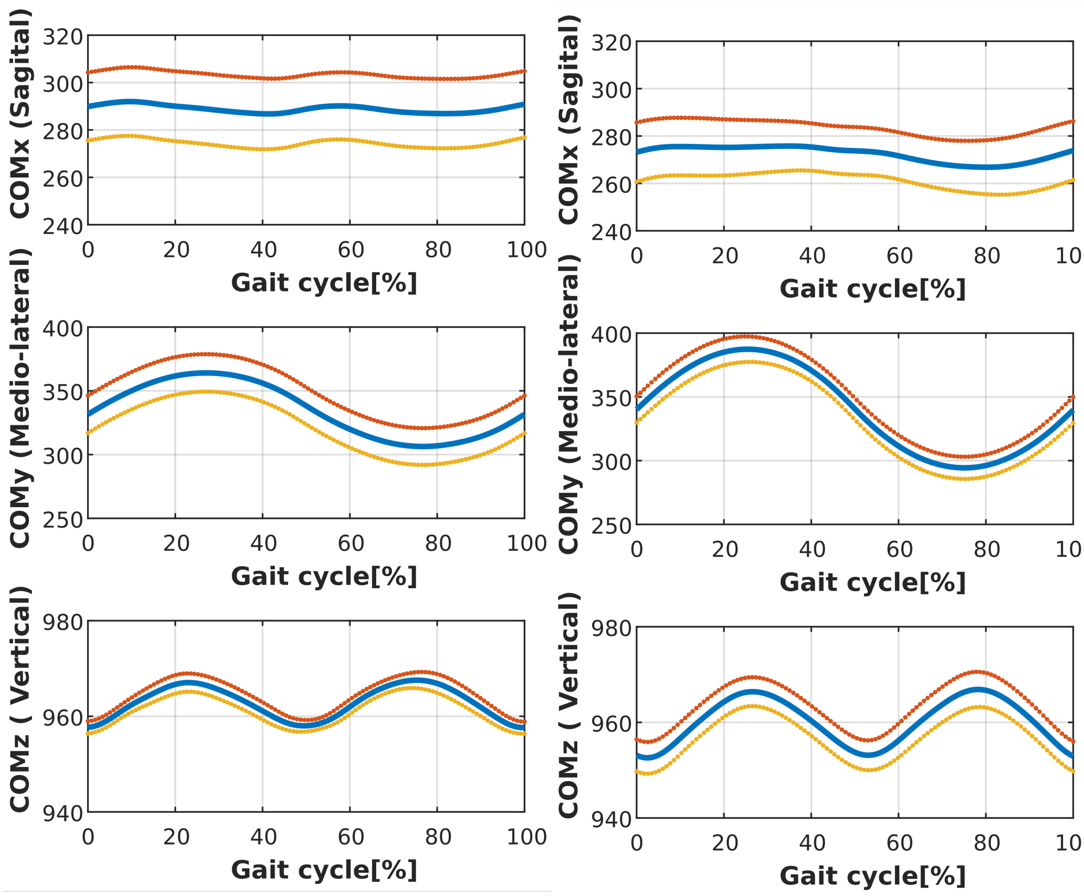


Figure 5.19: COM of the representative subject with unloading of 20%; left represent for BWS and right is for CWS. Red represents for COM value of mean + sd; yellow represents for COM value of mean – sd.

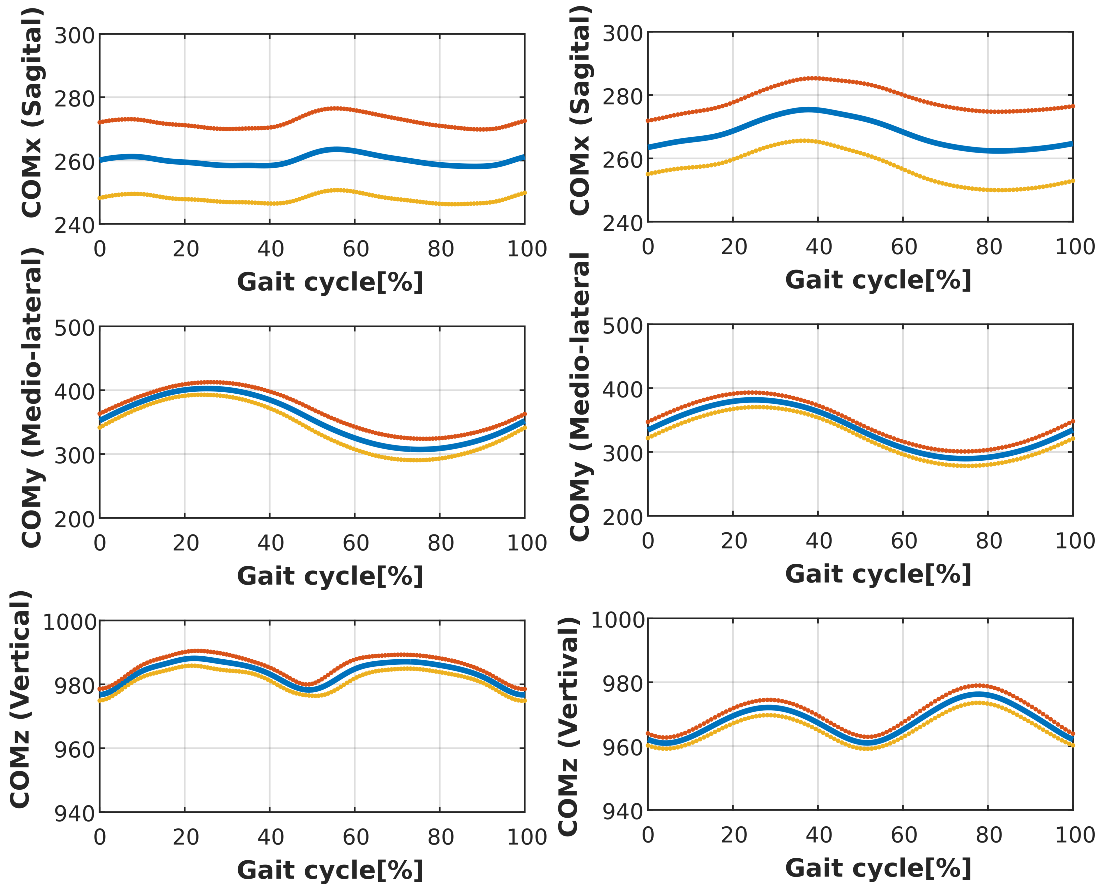


Figure 5.20: COM of the representative subject with unloading of 30%; left represent for BWS and right is for CWS. Red represents for COM value of mean + sd; yellow represents for COM value of mean – sd.

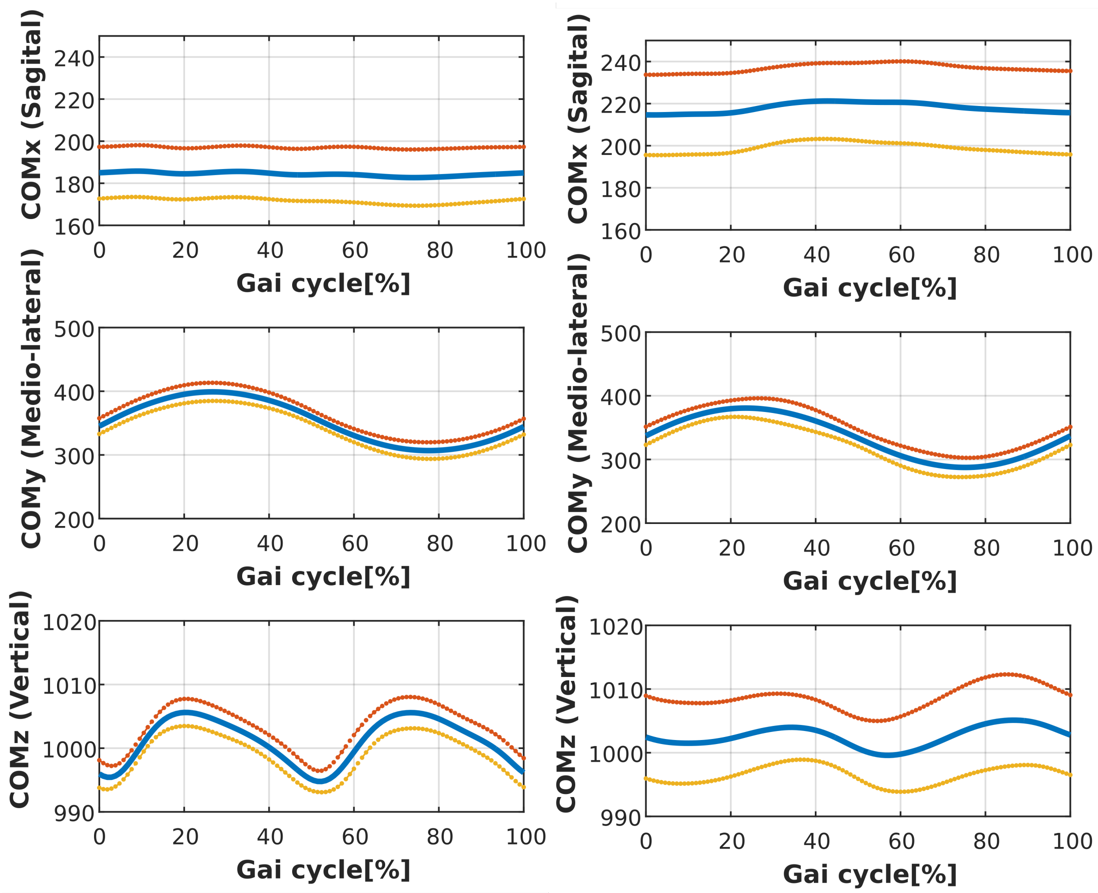


Figure 5.21: COM of the representative subject with unloading of 50%; left represent for BWS and right is for CWS. Red represents for COM value of mean + sd; yellow represents for COM value of mean – sd.

Table 5.3: Comparison COM in Y (Frontal) direction between BWS system, counter Weight system and Normal Walking of the representative subject by Paired t-test.

Method	Weight Support	Sample Size	Mean difference (N)	SD	SE mean	95% CI down	95% CI up
BWS	20%	2596	-25.535	32.704	0.642	-26.793	-24.276
CWS	20%	2596	-16.23	25.607	0.503	-17.216	-15.245
BWS	30%	2530	-6.2	25.327	0.504	-7.187	-5.212
CWS	30%	2530	-21.105	25.72	0.511	-22.107	-20.102
BWS	50%	2332	-4.655	28.202	0.584	-5.8	-3.509
CWS	50%	2332	-26.367	26.695	0.553	-27.451	-25.283

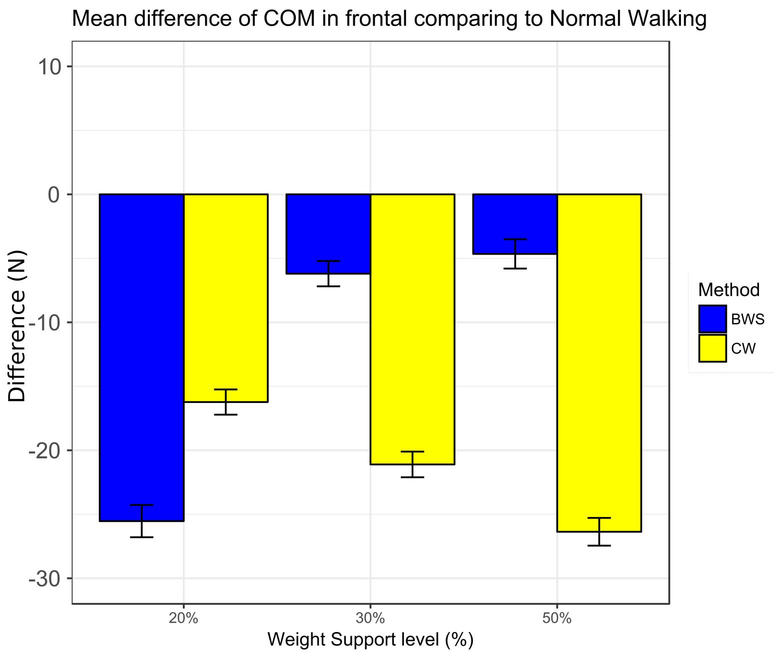


Figure 5.22: The demonstration of the difference of COM trajectories in frontal direction when using the weight support systems to COM trajectory in the frontal direction in in normal walking. Difference = COM by weight support system - COM normal walking.

In Figure 5.22, the difference of COM trajectories of the representative subject in frontal direction by using the weight support systems to the COM trajectory in normal walking is demonstrated. The more closed to 0 of the difference is the more similar of the COM trajectories by weight support system to the case of normal walking. The mean value is negative this represents the amplitude of COM trajectories since applying the weight support systems is smaller than in the case of normal walking. The difference of COM in the case of using Counter Weight system is increased when the weight support level increases. However, in the case of using the BWS system even is decreased follow the increasing of the weight support level. The limitation is due to the BWS system activates at two positions on subject unlike the case of the conventional weight support system. Then, when the weight support is more increased, this effect is clearer and makes the COM more similar to the case normal walking.

In order to compare COM trajectories in the frontal direction, the significant difference of the amplitude of COMy trajectories of all nine subjects is considered the BWS system and Counter Weight system as shown in Table 5.4. In Table 5.4, the COMy amplitudes for all subjects were standardized to the normal walking COMy mean value of each subject. The data that was standardized then was used for analyzing the difference of the COMy amplitude for both BWS system and Counter weight system using ANOVA. In Table 5.4, because of the standardization procedure, the mean value of the COMy in the normal walking case always equal to one, the closer to one of the mean value of the weight system cases is the more similar to the COMy of the normal walking case. From the Table 5.4, we may see that when the unloading force increases, the mean value of both BWS system and Counter weight system also decrease. On the Figure 5.23, the comparison of the mean value of BWS system and Counter weight system at difference weight support levels. We could see clearly that when increasing weight support level absolutely reduces the amplitude of COMy during walking. In the Figure 5.24, Figure 5.25 and Figure 5.26, the significant difference is found between the mean value of the normal walking case and both BWS system and Counter weight system ($p < 0.001$). These results strongly confirm that the COM in mediolateral is modified by using the unloading system. However, from the

Figure 5.24, Figure 5.25 and Figure 5.26, we could observe that the mean value COMy by using the BWS system is significantly higher than the case using the Counter Weight system ($p < 0.001$ at 30% weight support, and $p < 0.05$ at 50% weight support). We also see that the significant difference between the mean values is not found at the high level of weight support (at 70% weight support, $p = 0.927$). The closer mean value of the case using the BWS system is evidence such that the new system shows its better behavior to the Counter Weight system. The higher mean value of COMy in the case of applying the BWS system than Counter Weight system could be explained by the way that the unloading force that applies to the subject's trunk. The Figure 5.27 demonstrates the unloading force that applying on the subject' trunk for the Counter Weight system (subfigure A) and the BWS system (subfigure B). In the case of the Counter Weight system, the lateral part of unloading force tends to prevent the movement of the COM during walking and to pull the COM in mediolateral to the center axis. This affection is similar to the "pendulum effect" since in pendulum mechanism there always exists a lateral force. This "pendulum effect" may make the subject uncomfortable during walking and modify the gait parameters. However, in the case of the BWS system, the lateral part of the unloading force would be small because the unloading force in this case always tries to follow the moving of the COP during walking and reduces the effect of the lateral unloading force.

In Table 5.5, Consider the COM in Z direction of the representative subject we could see that the difference is increased when the weight support level is increased. Considering the Standard deviation of the difference between COM trajectories we could see the variance of the mean difference of COM by applying the BWS system for every level of body weight support are quite stable about 5(mm). The variance of the mean difference by applying the Counter Weight system is always higher than the case of applying BWS. At a high level of weight support, the variance by applying Counter Weight system is much greater than the case of using BWS system. This happens because of at the high level of weight support; the higher weight support levels will make the dynamic force is bigger due to the inertia of counter weights increasing. Moreover, this makes the subject uncomfortable during walking under high-level weight support of counter

Table 5.4: Comparison of the COMy amplitude in mediolateral of all nine subjects between BWS system, counter Weight system, and Normal Walking by ANOVA. NormalStd, BwsStd, and CwsStd are data of COMy amplitude standardized in cases of Normal walking, BWS system and Counter Weight system respectively.

Weight Support	Method	Sample Size	Mean	SD	95% CI
30%	NormalStd	123	1.0002	0.0841	(0.9716, 1.0288)
	BwsStd	123	0.8457	0.1809	(0.8170, 0.8743)
	CwsStd	123	0.7410	0.1957	(0.7124, 0.7696)
50%	NormalStd	127	0.9993	0.0953	(0.9712, 1.0274)
	BwsStd	127	0.6264	0.2114	(0.5984, 0.6545)
	CwsStd	127	0.5724	0.1548	(0.5443, 0.6005)
70%	NormalStd	122	0.9993	0.0838	(0.9715, 1.0271)
	BwsStd	122	0.3709	0.1929	(0.3431, 0.3987)
	CwsStd	122	0.3783	0.1693	(0.3506, 0.4061)

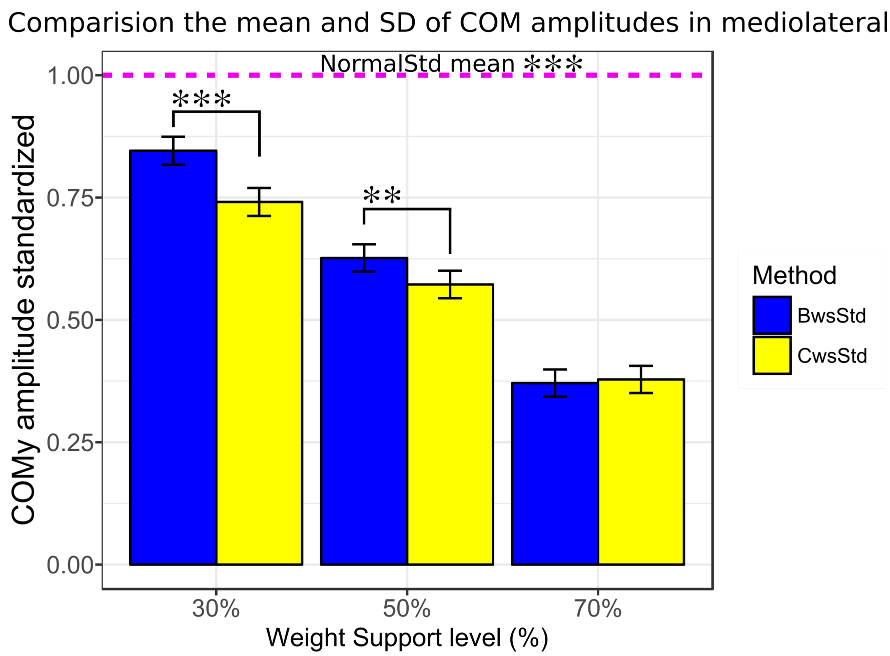


Figure 5.23: Demonstrate the comparison of the COMy amplitudes by using the BWS support system and Counter weight system at 30%, 50%, and 70% weight support.

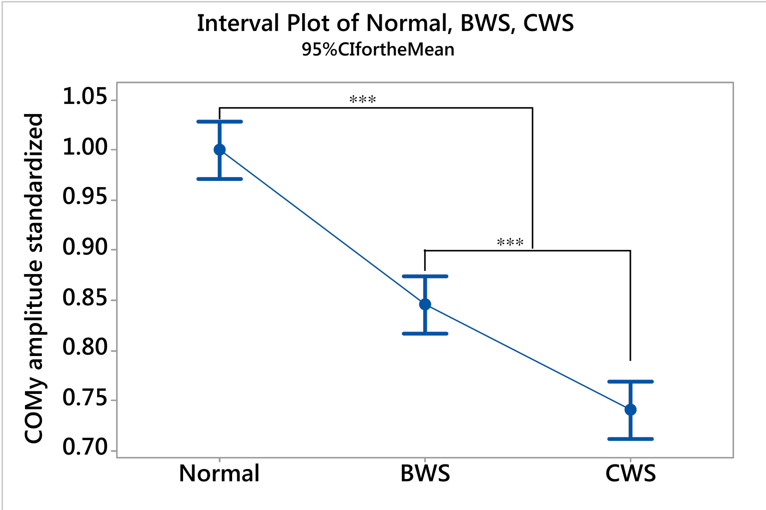


Figure 5.24: The demonstration of the difference of COMy amplitude (in mediolateral) of all subjects when using the weight support systems at 30% weight support, *** represented the significant value $p < 0.001$.

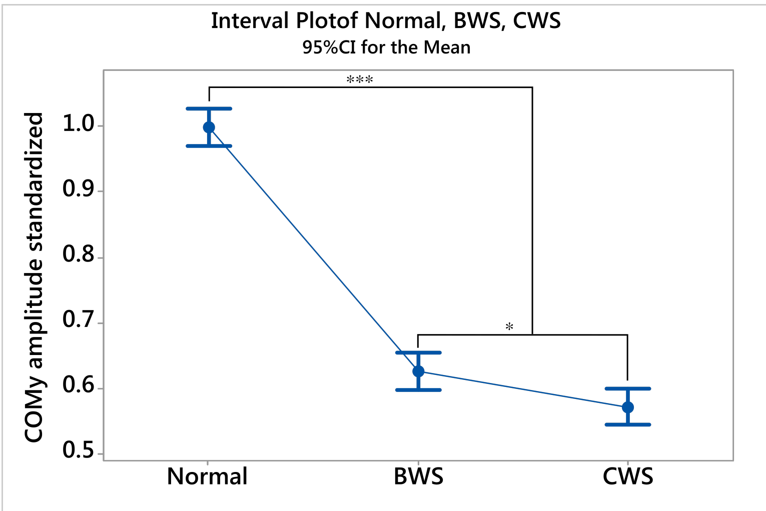


Figure 5.25: The demonstration of the difference of COMy amplitude (in mediolateral) of all subjects when using the weight support systems at 50% weight support; *** represented the significant value $p < 0.001$; * represented the significant value $p < 0.05$.

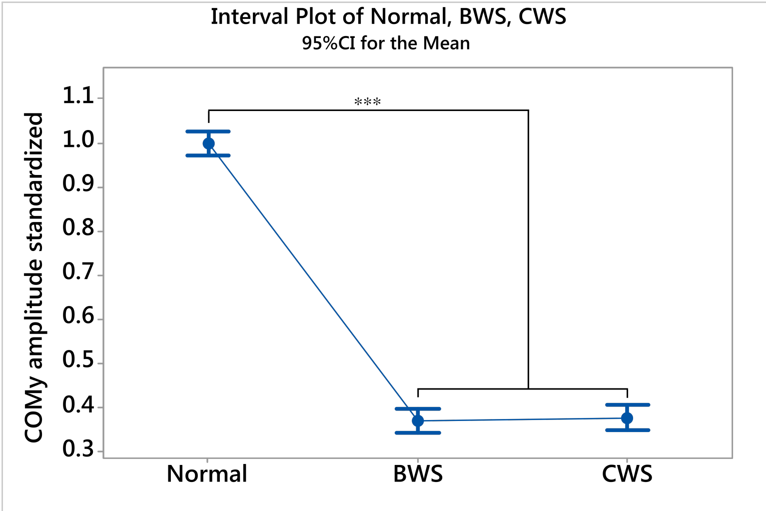


Figure 5.26: The demonstration of the difference of COMy amplitude (in mediolateral) of all subjects when using the weight support systems at 70% weight support, *** represented the significant value $p < 0.001$.

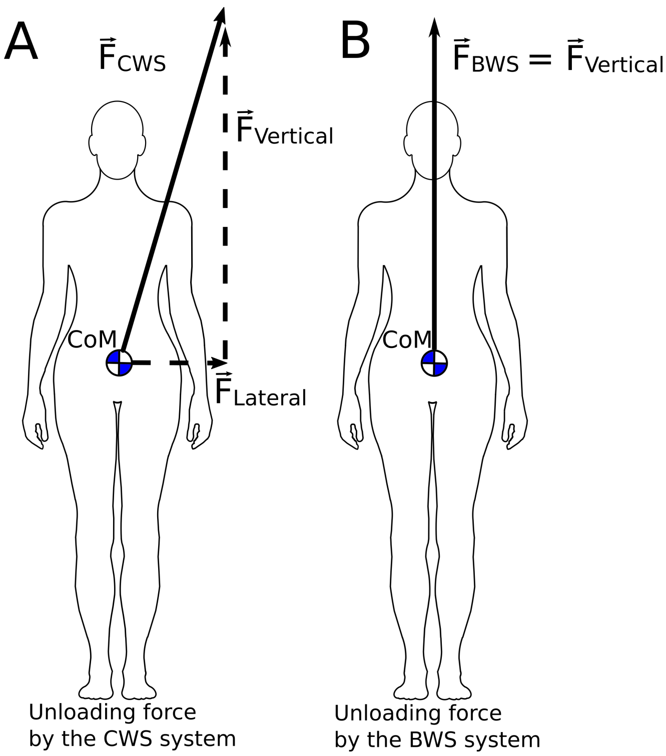


Figure 5.27: Demonstrate the unloading force applying to the subject's trunk.

Table 5.5: Comparison COM in Z (longitudinal) direction between BWS system, counter Weight system and Normal Walking of the representative subject by Paired t-test.

Method	**Weight Support**	**Sample Size**	**Mean difference (N)**	**SD**	**SE mean**	**95% CI down**	**95% CI up**
BWS	20%	2596	28.027	4.816	0.095	27.842	28.213
CWS	20%	2596	27.175	6.754	0.133	26.915	27.435
BWS	30%	2530	49.426	4.892	0.097	49.236	49.617
CWS	30%	2530	33.894	6.591	0.131	33.637	34.151
BWS	50%	2332	65.182	5.127	0.106	64.974	65.319
CWS	50%	2332	65.395	8.289	0.172	65.058	65.731

weight system. In contrast, the effect of dynamic force due to inertia in BWS system is absolute no, instead of, the unloading force generated by BWS system will adapt to the movement of the subject, then, the subject will feel comfortable as walking under BWS system.

Figure 5.28 demonstrates the variance of the differences in COM trajectories of the representative subject in the vertical direction when applying the weight support systems to the Com trajectory in normal walking. The vertical axis expresses the mean difference with standard deviation (SD), the horizontal represents the weight support level. The gray dot represents the BWS system, and the with dot represents the Counter Weight system. We may see that the segments of standard deviation are increased since the weight support levels increase. Moreover, at every weight support levels, the segment of SD by Counter Weight system is always longer than the case of using BWS system. In this result, we could strictly say that the COM trajectories in the vertical direction by using Counter Weight system are more variant than the case of using BWS system.

Table 5.6 represents the quantification of the COMz (in vertical) amplitudes of the normal walking, BWS system, and CWS system cases for all nine subjects. The data of COMz amplitude was also standardized similarly with the data of the COMy amplitude and analyzed by using ANOVA. In Table 5.6, we also

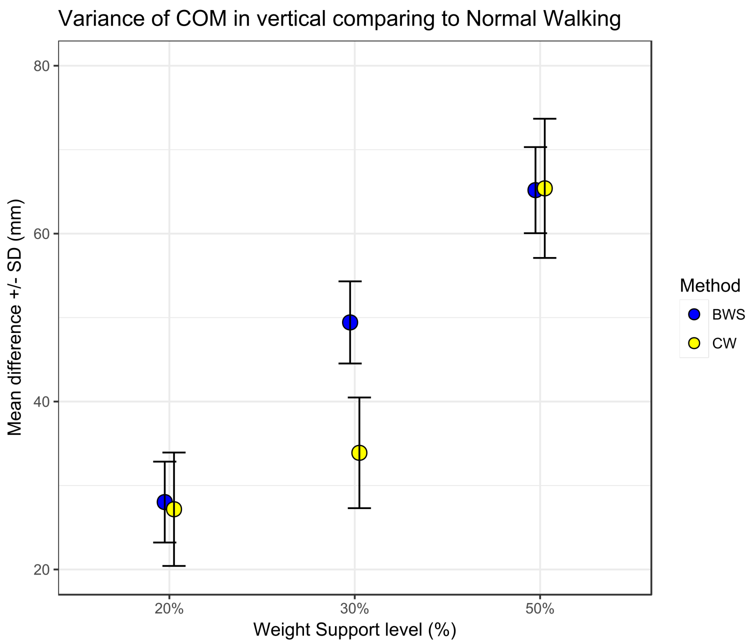


Figure 5.28: Demonstration the variance of differences in COM trajectories in the vertical direction of the representative subject when using the weight support systems to COM trajectory in the vertical direction in in normal walking. Difference = COM by weight support system - COM normal walking.

observe that the mean value of the normal walking case is around one. Meanwhile, all mean values of the weight support system case are smaller than the case of normal walking. The closing value of the mean value to one is the more similar amplitude of the COMz by using the weight support systems to the normal walking. The Figure 5.29 represents a comparison of the mean values for both BWS system and Counter weight system case. In the Figure 5.29, we could see that both BWS system and Counter Weight system reduced the amplitude of COMz. By increasing the weight support levels in case of using BWS system, the COMz amplitude slightly decrease. The significant difference of the mean can be observed only between the 30% weight support case and 70% weight support case ($p < 0.001$). There is no significant difference in the pair of 30% and 50% weight support ($p = 0.105$) and the pair of 50% and 70% weight support ($p = 0.169$). In the case of Counter Weight system, the significant difference of the mean cannot be found among cases 30%, 50%, and 70% weight support. However, one may see that at the high level of weight support the variance of the amplitude is extremely higher than the lower weight support level as well as the variance of the BWS system at all weight support level (around 0.41 compares to 0.18). The higher variance of COMz amplitude in the case of CWS system at the high level of weight support represented for the added oscillation of the COM in vertical to the normal COMz and this added oscillation may due to the stronger influence of "the pendulum effect" to the COM gait parameter. The Figure 5.30, Figure 5.31 and Figure 5.32 demonstrate the difference of the COMz amplitudes when using the weight supports system at 30%, 50% and 70% weight support levels, respectively. In these Figures, the significant difference of the mean value is found between the normal walking case and both BWS system and Counter weight system ($p < 0.001$). This results confirm that the COMz amplitude when subject used the unloading system is modified. However, one may see that the significant difference is found between the case of using the BWS system and Counter Weight system at the low and middle level of unloading force (30% and 50% weight support) ($p < 0.001$). Once again, this result confirms the better behavior of the BWS system than the Counter Weight system since the COM amplitude of BWS system closer to the normal walking. At the high level of unloading force (70% weight support), the mean value of the COMz amplitude in the case of using BWS system is slightly higher than the case using Counter Weight system. However, the significant difference of the COMz amplitude at the high level of weight support (70% weight support) is not found (p = 0.164). This result could be explained that the gait in locomotion at the high level of weight support is not unstable. Moreover, in the novel BWS system, the PMA that connect to the subject trunk is a soft actuator. And, at the strong pendulum influence to the locomotion gait, the moving virtual position of the unloading force may not compensate for the deformation of the COM movement.

Table 5.6: Comparison of the COMz (in vertical) amplitude of all nine subjects between BWS system, counter Weight system, and Normal Walking by using ANOVA. NormalStd, BwsStd, and CwsStd are data of COMy amplitude standardized in cases of Normal walking, BWS system, and Counter Weight system respectively.

Weight Support	**Method**	**Sample Size**	**Mean**	**SD**	**95% CI**
30%	NormalStd	144	1.0004	0.1207	(0.9785, 1.0224)
	BwsStd	144	0.7392	0.1163	(0.7172, 0.7612)
	CwsStd	144	0.6048	0.1610	(0.5829, 0.6268)
50%	NormalStd	143	1.0009	0.1116	(0.9748, 1.0272)
	BwsStd	143	0.7048	0.1704	(0.6786, 0.7310)
	CwsStd	143	0.5915	0.1866	(0.5653, 0.6177)
70%	NormalStd	143	1.0010	0.1209	(0.9573, 1.0447)
	BwsStd	143	0.6731	0.1545	(0.6294, 0.7168)
	CwsStd	143	0.6160	0.4166	(0.5723, 0.6597)

5.5 Discussion

The base of the experiment result performed with both two BWS system, the author could conclude that the novel Body Weight Support system successfully realizes weight support. Furthermore, one of the main points of the proposed BWS system is that the new system could follow the COP during subject walking

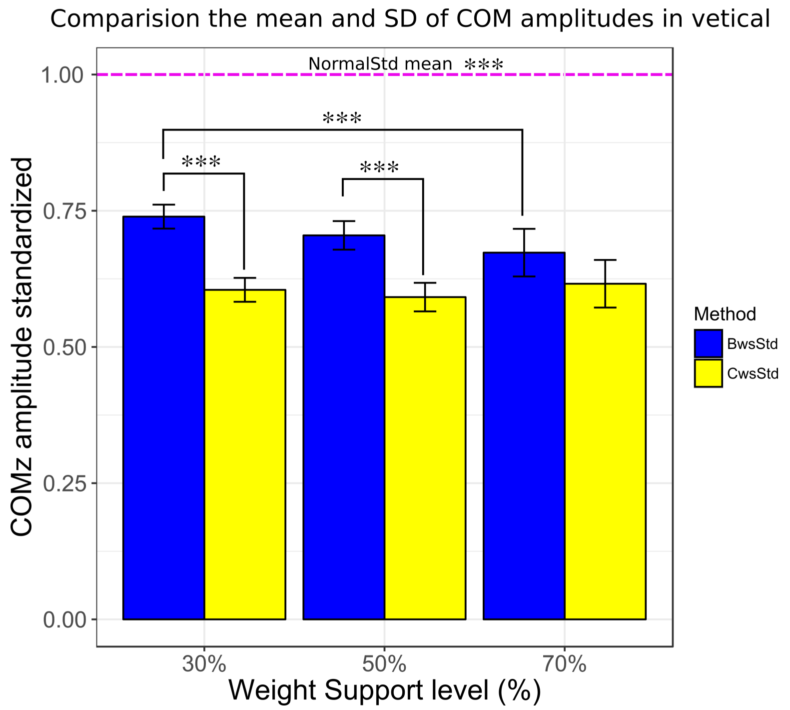


Figure 5.29: Demonstrate the comparison of the COMz amplitude by using the BWS support system and Counter weight system at 30%, 50%, and 70% weight support by using ANOVA. *** represented the significant value $p < 0.001$.

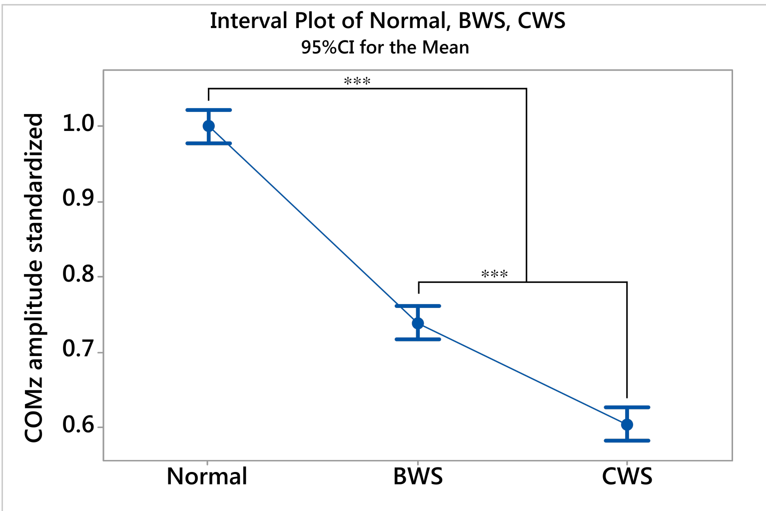


Figure 5.30: The demonstration of the difference of COMz amplitude (in vertical) when using the weight support systems at 30% weight support by using ANOVA, *** represented the significant value $p < 0.001$.

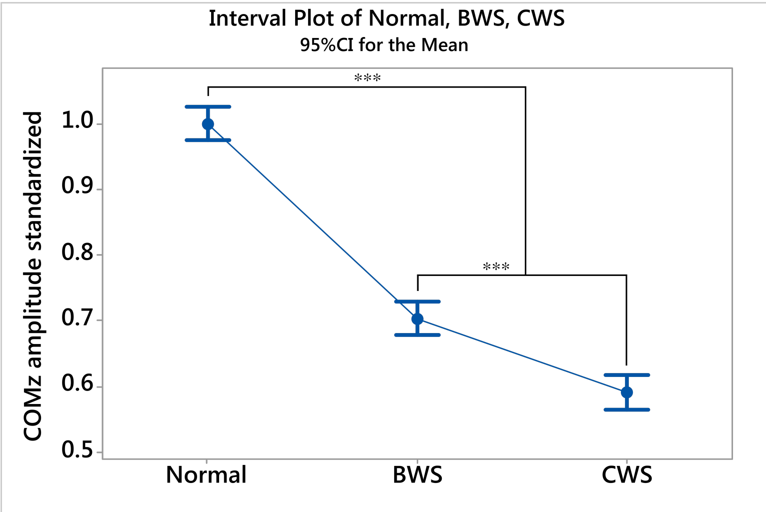


Figure 5.31: The demonstration of the difference of COMz amplitude (in vertical) when using the weight support systems at 50% weight support by using ANOVA, *** represented the significant value $p < 0.001$.

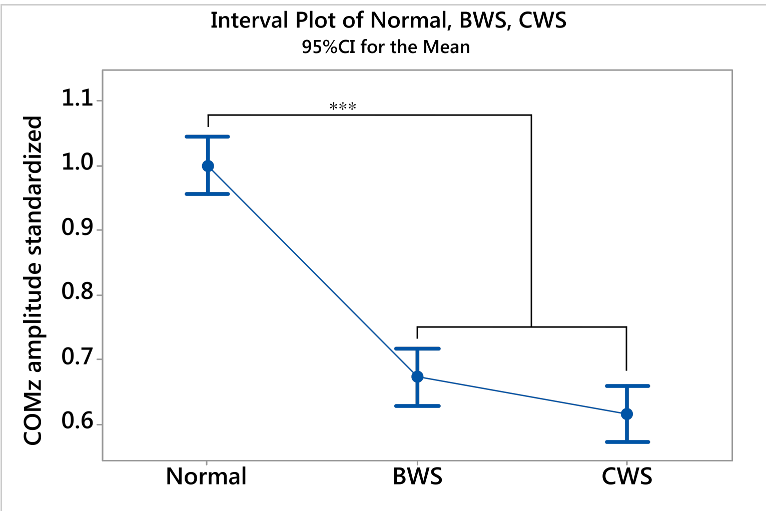


Figure 5.32: The demonstration of the difference of COMz amplitude (in vertical) when using the weight support systems at 70% weight support by using ANOVA, *** represented the significant value $p < 0.001$.

and provide the desired unloading force. Therefore, the generated unloading force by the new BWS system could make the subject more comfortable than the Counter Weight System. The fact that in almost experiments subject always confirmed that the new system makes the subject more comfortable than the Counter Weight system.

The first advantage of the novel BWS system is that it could easily tune the weight support level to get the desired unloading force. The new BWS system by using Pneumatic muscle actuators is active system then unloading force could easily be tuned by changing the pressure of the input compressed air. The Counter Weight System has itself limitation of tuning the weight support level, then the high level of unloading force it was difficult to catch up correctly. Then we could see that reaction force was a little bit higher than BWS system case, however, even though that the reaction force pattern generated by using Counter Weight system was still deformation. Furthermore, by using up to four muscle for BWS system, we could easily modulate the generated force for each side as we see in the Figure 5.10. Therefore, the total unloading loading force could change the virtual application point since the system tries to track the COP movement during the subject walking; and we also get the desired unloading force that follows the moving of COP. In Figure 5.10, we also could see the limitation of the control algorithm for the new system as the time delay for generated unloading force was still a bit high. However, the author believes that the new system still meets the basic criterion of a BWS system.

The second advantage is that the system prevents the "pendulum effect" during the walk that often happens when BWS system uses cable-pulley apparatus. Many common BWS systems use a rope as a connection between subject and an actuator (see [18], [10], [24], [17], [23]. The pendulum effect could be observed very clear in the result of the COM recorded by using the motion capture system and the quantification of the COM amplitude for every subject. The COM patterns by using the BWS system were also stable and less variant that was represented by the standard deviation error for every gait cycle recorded. On the contrary, the COM paths by using the Counter Weight system were much changed and variant. Especially, at the high level of weight support, the movement of the

COM look like has an addition fluctuation along every gait cycle. The result from the COM amplitude quantification for all subjects also confirmed that the variance of the mean COM amplitude in the case using Counter Weight system is always higher the case using BWS system, especially, it was extreme higher at the high level of weight support (70% weight support). The reason for this fluctuation added to the COM path could be explained by the "pendulum effect" which often happened at the conventional body weight support system using the pulley-rope apparatus. Because of the "pendulum effect," the lateral part of unloading force tends to pull the COM to the center axis. Therefore, the COM in mediolateral applying the weight support system, in general, has a smaller amplitude in comparison with the normal walking. However, the influence of the lateral force in conventional body weight support system is stronger than the new BWS system so that the COM amplitude is always smaller. The investigation of the COM amplitude confirms the better behavior of the new BWS system in comparison with the Counter Weight system. The higher variance of the COM amplitude in vertical happened due to the dynamic character of counterweight, this counter weights used for unloading system was always moving like an oscillation during walking. This oscillation added a bit more force to the unloading force, therefore, the effect to the COM moving. The new BWS system using pneumatic muscles then this affection was removed then this makes the gait training totally more comfortable for the subject.

The other advantage of BWS system using PAMs is that the unloading force directly acts on subjects and does not use any intermediary apparatus. Using PAMs makes the system very simple and easy to develop and improve and it also easy to apply the suitable the therapy strategy based on weight support.

5.6 Conclusion

In this chapter, the method for data collection was discussed. Every experiment results which included reaction forces, unloading forces, COP trajectories and COM trajectories and the quantification of the step width, COM amplitude for both the BWS system and the Counter Weight System were presented. The analyzing of the effects the BWS system and Counter Weight system were conducted and discussed carefully. From the assessment, the new BWS system represents the better behavior than the Counter weight system. Even the news system still has itself some limitations. However, it has many advantages in comparison, particularly, with the conventional pulley-rope weight support system type.

Chapter 6

Conclusion and Future work

This dissertation successfully proposed a novel active BWS system using pneumatic muscle actuators for rehabilitation system for gait training. Many types of body weight systems were reviewed to cover the advantages and disadvantages of each type.

From the new design for the new BWS system, the dissertation proposes a totally new way for implement the unloading for the rehabilitation of patient on the treadmill. The author strongly convinced that the common static BWS systems do not enough provide much comfort to the subject and such that give good results on the rehabilitation of patients.

For every main problem of this research such as designing, controlling, validating and assessment for active BWS system was also discussed very carefully to get the best solution for the new BWS system. For designing, the author reviewed many types of previous design and proposed numerous of apparatus for Body Weigh Support system. For the controlling, the dissertation also discussed quite details of the selection. For validating and assessment, the experiments were implemented and compared two different systems, a classic one and new one based on the utilization of PAMs. The main difference between the systems based on PAMs is just the way that the unloading force is generated. In the active BWS system, instead, we use the COP tracking model to provide input to give the patient the freedom to oscillate as during a normal walk. The data were recorded the reaction force, COP path and the COM path of the subject in the implemented systems for comparison the new system and counter weight system. The results show that the new system is the best in reproducing the behavior of a normal walk and has much advantage than the counterweight system or the conventional pulley-rope BWS system.

There are some limitations of the Body Weight System using pneumatic muscle actuator. The pneumatic muscle using for this BWS system has its complex character such as nonlinear and hysteresis behavior which lead to difficulties in control system. Then the result is that it would be difficult to get the desired unloading precisely tracking the reference unloading force as seen in Figure 5.10. However, even though that the new system is still better than the counterweight system and the author believes that the new BWS system is still met the criterion for weight support system.

For the future work will focus on giving more freedom to the patient especially by also considering the oscillation on the y-axis to get full COP tracking model. The control strategy for the pneumatic muscle will also consider getting the most precise unloading force tracking which could make the results even better.

References

[1] Combs. S. A., Dugan. E. L., Ozimek. E. N., and Curtis. A. B. Effects of body-weight supported treadmill training on kinetic symmetry in persons with chronic stroke. *Clin Biomech Bristol Avon*, **27**(9):887–92, July 2012.

[2] Grecco. L. A., Tomita. S. M., Christovao. T. C., Pasini. H., Sampaio. L. M., and Oliveira. C. S. Effect of treadmill gait training on static and functional balance in children with cerebral palsy: a randomized controlled trial. *Brazilian journal of physical therapy*, **17**(1):17–23, March 2013.

[3] Kyvelidou. A., Kurz. M. J., Ehlers. J. L., and Stergiou N. Aging and partial body weight support affects gait variability. *J Neuroeng Rehabil*, **5**(1):22, Sept 2008.

[4] G. Andrikopoulos, G. Nikolakopoulos, I. Arvanitakis, and S. Manesis. Piecewise affine modeling and constrained optimal control for a pneumatic artificial muscle. *Industrial Electronics IEEE Transactions*, **61**(2):904–916, Mar 2014.

[5] Dobkin. B, Apple. D, Barbeau. H, Basso. M, Behrman. A, Deforge. D., Ditunno. J., Dudley. G., Elashoff. R., Fugate. L, Harkema. S., Saulino. M., and Scott. M. Weight-supported treadmill vs over-ground training for walking after acute incomplete sci. *Neurology*, **66**(4):484–93, Feb 28 2006.

[6] Tondu B. and P. Lopez. Modeling and control of mckibben artificial muscle robot actuators. *Control Systems IEEE*, **20**(2):15–38, Apr 2000.

[7] Chou C.-P. and B. Hannaford. Measurement and modeling of mckibben pneumatic artificial muscles. *Robotics and Automation IEEE Transactions*, **12**(1):90–102, Feb 1996.

[8] Lewek. M. D. The influence of body weight support on ankle mechanics during treadmill walking. *Journal of biomechanics*, **44**(1):128–33, Jan 04 2011.

[9] Caldwell D.G., G.A. Medrano-Cerda, and M. Goodwin. Control of pneumatic muscle actuators. *Control Systems IEEE*, **15**(1):40–48, Feb 1995.

[10] Gazzani F, Fadda A, Torre M, and Macellari V. Ward: a pneumatic system for body weight relief in gait rehabilitation. *IEEE Transactions on Rehabilitation Engineering*, **8**(4):506–13, Dec 2000.

[11] Prattico F., M.A.M. Dzahir, , and S. i. Yamamoto. Computed-torque method for the control of a 2 dof orthosis actuated through pneumatic artificial muscles: a specific case for the rehabilitation of the lower limb. *arXiv preprint*, **arXiv:1404.6968**, 2014.

[12] Andrikopoulos G., G. Nikolakopoulos, and S. Manesis. Advanced nonlinear pid-based antagonistic control for pneumatic muscle actuators. *Industrial Electronics IEEE Transactions*, **61**(12):6926–6937, Dec 2014.

[13] Colombo G., Joerg M.and Schreier R., and Dietz V. Treadmill training of paraplegic patients using a robotic orthosis. *Journal of rehabilitation research and development*, **37**(6):693–700, Nov-Dec 2000.

[14] Lovely. R. G., Gregor. R. J., Roy. R. R., and Edgerton. V. R. Effects of training on the recovery of full-weight-bearing stepping in the adult spinal cat. *Experimental neurology*, **92**(2):421–35, May 1986.

[15] Aschemann H. and D. Schindele. Comparison of model-based approaches to the compensation of hysteresis in the force characteristic of pneumatic muscles. *Industrial Electronics IEEE Transactions*, **61**(4):3620–3629, 2014.

[16] Barbeau. H. and S. Rossignol. Recovery of locomotion after chronic spinalization in the adult cat. *Brain research*, **412**(1):84–95, May 1987.

[17] Barbeau H., M. Wainberg, and L. Finch. Description and application of a system for locomotor rehabilitation. *Medical and Biological Engineering and Computing*, **25**(3):341–344, May 1987.

[18] He J., R. Kram, and T.A. McMAHON. Mechanics of running under simulated low gravity. *Journal of Applied Physiology*, **71**(3):863–870, Sep 1991.

[19] Nymark. J, DeForge. D, Barbeau. H, Badour. M, Bercovitch.S, Tomas.J, Goudreau.L, and MacDonald. J. Body weight support treadmill gait training in the subacute recovery phase of incomplete spinal cord injury. *Journal of Neurologic Rehabilitation*, **12**(3):119–136, Sept 1998.

[20] Lilly J.H. Adaptive tracking for pneumatic muscle actuators in bicep and tricep configurations. *Neural Systems and Rehabilitation Engineering IEEE Transactions*, **11**(3):333–339, Sept 2003.

[21] Lilly J.H. and P.M. Quesada. A two-input sliding-mode controller for a planar arm actuated by four pneumatic muscle groups. *Neural Systems and Rehabilitation Engineering IEEE Transactions*, **12**(3):349–359, Sept 2004.

[22] Finch. L., H. Barbeau, and B. Arsenault. Influence of body weight support on normal human gait: development of a gait retraining strategy. *Physical Therapy*, **71**:842–855, Nov. 1991.

[23] Qi Lu, Jianxun Liang, Bing Qiao, and Ou Ma. A new active body weight support system capable of virtually offloading partial body mass. *IEEE/ASME Transactions on Mechatronics*, **18**(1):11–20, Feb 2013.

[24] Frey M, Colombo G, Vaglio M, Bucher R, Jörg M, and Riener R. A novel mechatronic body weight support system. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, **14**(3):311–321, Sep 2006.

[25] Glauser M., Z. Lin, and P.E. Allaire. Modeling and control of a partial body weight support system: an output regulation approach. *IEEE Transactions on Control Systems Technology*, **18**(2):480–490, Mar 2010.

[26] Visintin. M., H. Barbeau, and Korner-Bitensky. N.and Mayo. N. E. A new approach to retrain gait in stroke patients through body weight support and treadmill stimulation. *Stroke*, **29**(6):1122–8, Jun 1998.

[27] Kosak. M.C. and M.J. Reding. Comparison of partial body weight-supported treadmill gait training versus aggressive bracing assisted walking post stroke. *Neurorehabil Neural Repair*, **14**(1):13–9, March 2001.

[28] Adams. M.M. and A.L. Hicks. Comparison of the effects of body-weight-supported treadmill training and tilt-table standing on spasticity in individuals with chronic spinal cord injury. *The journal of spinal cord medicine*, **34**(5):488–494, Nov 2011.

[29] Franz J. R., Riley P. O., Dicharry J., Allaire P. E., and Kerrigan D. C. Gait synchronized force modulation during the stance period of one limb achieved by an active partial body weight support system. *J Biomech*, **41**(15):3116–20, Nov 2008.

[30] Kram R., A. Domingo, and D.P. Ferris. Effect of reduced gravity on the preferred walk-run transition speed. *Journal of Experimental Biology*, **200**(4):821–826, Feb 1997.

[31] R. M. Robinson, C. S. Kothera, R. M. Sanner, and N. M. Wereley. Nonlinear control of robotic manipulators driven by pneumatic artificial muscles. *IEEE/ASME Transactions on Mechatronics*, **21**(1):55–68, Feb 2016.

[32] Gravano. S., Ivanenko. Y. P., Maccioni G., Macellari V., Poppele. R. E., and Lacquaniti. F. A novel approach to mechanical foot stimulation during human locomotion under body weight support. *Hum Mov Sci*, **30**(2):352–67, Apr 2011.

[33] Hesse. S., M. Konrad, and D. Uhlenbrock. Treadmill walking with partial body weight support versus floor walking in hemiparetic subjects. *Archives of Physical Medicine and Rehabilitation*, **80**(4):421–7, Apr 1999.

[34] Van Thuc T., F. Prattico, and S. i. Yamamoto. A novel treadmill body weight support system using pneumatic artificial muscle actuators: a comparison between active body weight support system and counter weight system. In *World Congress on Medical Physics and Biomedical Engineering*, pages 1115–1119, June 2015.

[35] Hornby T.G., D.H. Zemon, and D. Campbell. Robotic-assisted, body-weight–supported treadmill training in individuals following motor incomplete spinal cord injury. *Physical therapy*, **85**(1):52–66, Jan 2005.

[36] Dietz V., R. Müller, and G. Colombo. Treadmill training of paraplegic patients using a robotic orthosis. *Brain*, **125**(12):2626–34, Dec 2002.

[37] Dietz V., R. Müller, and G. Colombo. Treadmill walking with body weight support: Effect of treadmill, harness and body weight support systems. *Gait and posture*, **28**(2):303–308, Mar 2008.

[38] T. Vo-Minh, T. Tjahjowidodo, H. Ramon, and H. Van Brussel. A new approach to modeling hysteresis in a pneumatic artificial muscle using the maxwell-slip model. *EEE/ASME Transactions on Mechatronics*, **16**(1):177–186, Feb 2011.

List of Publications

[P.1] T.V. Thuc, F. Prattico, M.A.M. Dzahir, and S. Yamamoto, "A novel Treadmill Body Weight Support System using Pneumatic Muscle Actuators," *EMBC 2014*, Chicago, USA, 27-30/08/2014.

[P.2] Tran Van Thuc, Flavio Prattico, and Shin-ichiroh Yamamoto, "A novel Treadmill Body Weight Support system using Pneumatic Artificial Muscle actuators: a comparison between active Body Weight Support system and counter weight system," *World Congress on Medical Physics and Biomedical Engineering*, June 7-12, 2015, Toronto, Canada Volume 51 of the series IFMBE Proceedings pp 1115-1119.

[P.3] Tran Van Thuc and Shin-ichiroh Yamamoto, "Development of a body weight support system using pneumatic muscle actuators: Controlling and validation," *Advances in Mechanical Engineering*, Vol 8, Issue 12, page 1687814016 68359, Dec 2016. DOI: 10.1177/1687814016683598.

[P.4] 滝口理一,Tran Van Thuc,大島達也,川上拓真,野口洋平,飯村仁一,萩原杜子,柴田芳幸,山本紳一郎 ～ 空気圧人工筋を用いた免荷式歩行訓練システムの開発～免荷装置による歩行姿勢制御と評価 ～ ライフサポート学会フロンティア講演会, 2017年3月.