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下半身輔具及防跌倒機構之設計

Design of walking-assistive device on lower limb with fall-preventing mechanism

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中文摘要

近年來，高齡化的現象越來越明顯，年長者的比例大幅度的上升，而防止老人跌倒的問題就成了一個很大的議題，在年長者中有將近三分之一的人曾發生過跌倒，然而，這情況發生後發生第二次跌倒的機會更大，在第一次跌倒後，需要耗費大量的人力以及社會成本來照顧以及避免再次的跌倒。人們常說預防勝於治療，本論文希望可以藉由下半身外骨骼輔具的機構設計，在特定的步態下，將小腿以及大腿的相對轉動固定，期望能夠達到防止跌倒的功能，藉此減少因老人跌倒而造成的人力及社會成本。

首先，為了不讓老人穿上下半身外骨骼後，反而造成其身體不方便，所以，我們希望外骨骼的機構可以很適合人體，我們探討下半身的兩的關節，膝蓋關節以及髖關節，膝蓋關節是一個鉸鏈關節，除了轉動以外，膝蓋關節還有滑動的動作，然而，膝蓋是一個包括前後十字韌帶以及多條肌肉連動所形成的結構，為了避免下肢外骨骼輔具設計太過複雜以及過於笨重，膝蓋的部份我們將其簡化成只有鉸鏈關節。然而，不同於其他的下肢輔具，我們希望可以把原有大部分下肢輔具中，膝蓋關節部分的馬達用簡單機構的方式取代，以此降低整體輔具的重量。然後，髖關節的部分是球窩關節，擁有屈伸以及內收外展的自由度，但其轉動軸心在身體內部，以穿著在外的輔具來說，則是將兩個自由度分別拆開，並且將兩個自由旋轉軸心交於髖關節的球窩中心，以此達到類似髖關節的運動方式，而髖關節的部分，雖然有兩個自由度，我們只考慮使用馬達控制屈伸的自由度，以此來抬起外骨骼輔具，至於內收外展的自由度，目的是為了使穿戴著不會因此外骨骼而限制其原本習慣的走路狀態，所以將其變成被動的轉軸，當行走的時候他會隨者人而自己轉動。

再來，我們有了外骨骼的雛形後，再分析步態的細節，找出可以防止跌倒的可能性，我們在小腿及大腿發生相對轉動的膝蓋上，設計出一個滑軌以及滑塊的機構，藉此可以達到在單腳站立的步態時，限制膝蓋關節的轉動，達到固定且防止跌倒的效果，但這個固定卻不會使腳在擺動步態的狀態時，限制住大腿及小腿的相對運動，換言之，藉由此機構，達到需要固定時固定之，需要使膝蓋關節轉動時，讓它轉動。

最後藉由向量迴路法，分析出適合的尺寸及大小，並且先以木頭製造出簡單的模型，再將此模型大幅度縮小及美化，製造出最後的模型。實驗部分，希望可以由外

骨骼的角度變化，配對其正常人的角度變化，看是否此外骨骼的設計可以貼近實際人類的行走狀態。而在於防跌倒機構的部份，可知機構已達到我們期望的效果，但是至於此效果可否達到防止跌倒的功能是否，目前還沒找到驗證的方法，未來會尋找一個適合驗證的方案。

關鍵字：外骨骼；步態分析；防止跌倒

ABSTRACT

The population of the elders has dramatically increased, and fall prevention has become an important issue for elders care. This paper attempts to design a lower-limb exoskeleton, which could fix the relative rotation of the thigh and calf at specific gait to make the function of fall prevention and helping elder's walk. Unlike other assist devices, we replace the motor by special mechanism at knee joint for reducing the weight and making control easier on the device. We separate hip joint into two axes (rotation of flexion/extension and adduction/abduction) on the assistive device and make these two axes intersect at the center of the hip joint. Therefore, we can make the joint on the assistive device similar to hip joint. Based on human gait, we design a slider mechanism on the knee part that could be fixed for the leg at its state. Finally, we optimize the design and test a metal prototype. Through experiments, we measure the rotation about hip and knee to verify the function of our assistive device. About the fall-preventing mechanism, we can know the mechanism can reach our purpose, but we don't know whether this method can reach fall-preventing. In future, we will find some methods to prove the fall-preventing.

Keywords: exoskeleton; gait analysis; prevent of falls

SYMBOL TABLE

r_1	Length between hip joint to knee joint
r_2	Length between the rotation center of knee to the slider center
r_3	Change of the slider movement in the sliding groove
r_4	Length of the fixed bar
r_5	Length from motor shaft to the center of the fixed end
θ_1	Angle between r_1 and level line.
θ_2	Angle between r_2 and level line.
θ_3	Angle between r_3 and level line.
θ_4	Angle between r_4 and level line.
θ_5	Angle between r_5 and level line.
α	Rotation degree of hip joint
β	Rotation degree of knee joint

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Chapter 1 Introduction

1.1 Background

1.1 Background

In recent years, the population of elders was greatly increasing day by day. The number of elderly people in the world has been consistently and proportionally increasing. In 2010, there were 440 million people in the world aged above 65 years in total [1], and will be forecasted to 1,555 million by 2050 [2] . Therefore, problems of the elderly care gradually show up. If the elderly have the ability to take care of themselves in their daily lives such as walking to toilet or taking a walk, the manpower to take care of the elderly can be reduced it also can solve the problem of care for the elderly. However, the falling issue of elders is a common and important problem. There would be more problems after the elders fell down. Therefore, preventing falling is an important issue. There are various risk factors which cause fallings of elders as listed in Table 1.1[3-8]

Many serious falls cause fractures and may lead to head injuries, depression, fear of recurrent falls, gait abnormality, sensor impairment, cognitive impairment, and social isolation. These may cause great social burden and make it more difficult to take care of the elderly. Most of the methods to prevent falling are in line with problems, such as treatment by medications, sort out the environment, etc. In addition to preventive measures above, rehabilitation gait training is a method, too. In this paper, we hope to design a mechanism to make the elderly walk normally and prevent falling in order to reach the ideal that the elderly can take care of themselves and walk safely.

Table 1-1 Risk factor of falling

Physical Factors	● Reduced gait and balance. ● Functional declination: brain, joint and muscles, leading to uncoordinated movement. ● Reduced sensory functions: visual, auditory, tactile. ● Active of daily living inclined: exhaustive or too little physical exercise, improper dress and uncomfortable shoes.
Psychological factors	● Fear of fall: a history of falls leading to self-constraints ● Bad mood: caused by pain of disease, financial stress, sleep disturbance, etc.
Environmental factors	● Cleanliness of environment, lighting, bed height, toilet handles, slippery and uneven surfaces, use of walking helpers, etc.
Disease factors	● Neurology: stroke, Parkinson’s disease, posture and movement disorders, cerebellar disorder, vestibular disorder, etc. ● Metal disease: depression, sleep disorder, delirium, etc.

1.2 Overview of walking assist device

Walking assist device can be traced from the history of exoskeleton. Figure 1.1 shows the first set of exoskeleton –Hardiman, which was manufactured by GE in 1966 [9]. Exoskeleton is a multifunction assist device. According to different types, assist device has different benefits. Generally, the purpose of assist device can be divided into enhancement and assistance [10]. It can also be divided into four purposes for assistance, enhancement, protection, and detection [11]. The program of assistance enabled individuals with spinal cord injuries or handicapped to stand and walk again. Next the enhancement program improved the operator's abilities, such as taking heavier objects, running faster or endurance augmentation. And the protection program had different protection functions which depended on the design of mechanism and material. It can also be used in the safety of nuclear power plant or protect the work during walking. As for the detection program, it obtained the information from human body with sensors. The information was used to benefit rehabilitation and treatment. Moreover, the information can be used to control the exoskeleton. In the recent years, the researches of the exoskeleton are always combined with the application of above four purposes. So the exoskeleton can be divided into two categories, and described in the following article.

Figure 1.1 Hardiman

1.2.1 Exoskeleton of enhancement

In recent decades, the study of the exoskeleton had multiple functions, so it can be divided into two categories: enhancements, and assist. First, we introduce the enhancement type of exoskeleton.

Figure 1.2 shows the exoskeleton BLEEX which is developed by Kazerooni and his research team in 2004 [12]. In order to conform to ergonomics, BLEEX comprises two powered anthropomorphic legs, a power supply and a backpack-like frame on which a variety of heavy payloads can be mounted. BLEEX also use a large number of sensors and hydraulic actuators. And the combination with inclinometer judging makes BLEEX be close to a human control system. In the other hand, BLEEX can reduce the wearer’s burden, wearer can walk without feeling any load at walking speed below 3.2 kilometers per hour.

Next, Kazerooni and his research team strengthened the capacity of their exoskeleton. Figure 1.3(a) shows ExoHiker^{TM} [13] which was completed in February 2005. With the weight of 14kg including power unit, batteries and on-board computer, ExoHiker^{TM} can make the wearer feel no load with 70 kg mounted. Compared with BLEEX, ExoHiker^{TM} increased the loads and reduced the weight. It also had a solar panel and an 80 W-hour battery so that ExoHiker^{TM} can carry heavy loads during long mission. Generally speaking, ExoHiker^{TM} can carry 70 kg for 21 hours while the wearer feels no load on his shoulder. In the same year, ExoClimber^{TM} [14] retained the same long term load carrying capability of ExoHiker™, as shown in Figure 1.3(b). Although its weight was 22 kg, it was design to allow rapid stair ascent. The mission range possesses at least 200 mm ascent per 500g of battery while carrying 70kg payload. Without a doubt, ExoClimber^{TM} can also carry 70 kg of load as the wearer feeling no load on his shoulder. Figure 1.3(c) shows UCLC^{TM}（Human Universal Load Carrier）which was combined with ExoHiker™ and

ExoClimber™. UCLC™ enhanced the capacity, it can carry 90 kg. In particular, the oxygen consumption of these users carrying a 36 kg approach load at a speed of 3.2 km per hour decreased by about 15% when using the prototype HULC™. The function of reducing burden on the user is a great development, reducing burden for long-term task is very important for soldiers [15].

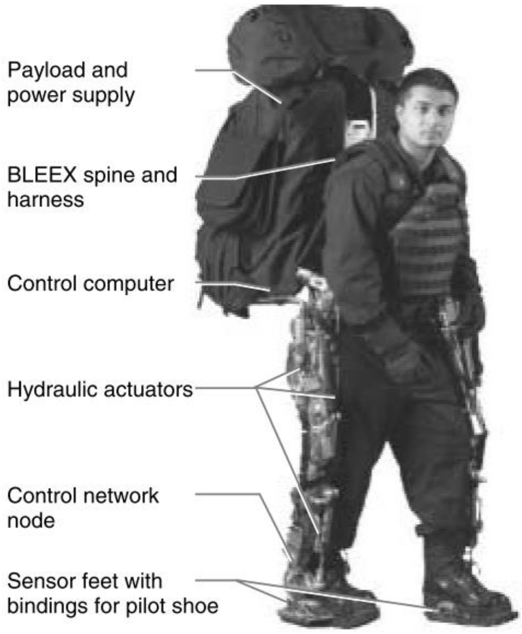


Figure 1.2 Berkeley Lower Extremity Exoskeleton (BLEEX)

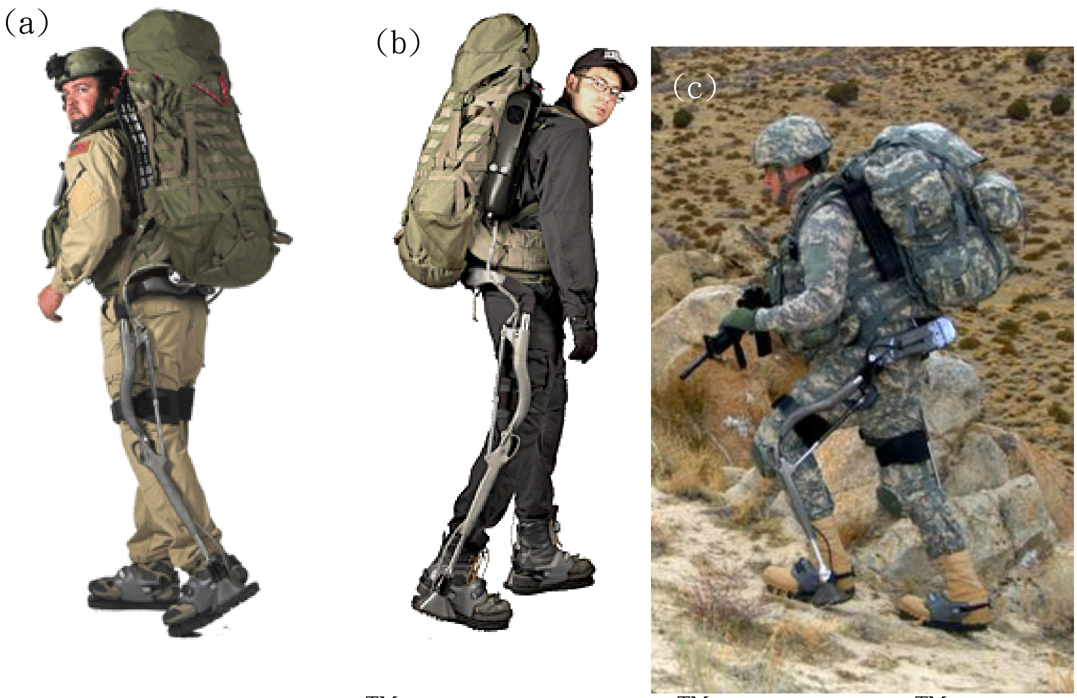


Figure 1.3 (a) ExoHiker^{TM} [14]. (b) ExoClimber^{TM} [15]. (c) UCLC^{TM} [16].

1.2.2 Exoskeleton of assist

Then, we introduce the assist type of exoskeleton. In this paper, our purpose is to design a lower limb assist device, the purpose of our assist device is similar to this type of exoskeleton. There are many functions in assist exoskeleton. For example, Honda developed the Walking Assist Device programs [16,17], University of Tsukuba developed Hybrid Assistive Limb [18], and Argo Medical Technologies developed the ReWalk which was the only exoskeleton system passed FDA (Food and Drug Administration).

Figure 1.4 shows the appearances of Honda's Walking Assist Device. Honda didn't call the device "exoskeleton," they called it walking assist device since the purpose of this programs are assisting in walking not like armor wear on body. Figure 1.4(a) shows Walking Assist Device with Stride Management Assist (WADSM) which is a lightweight (about 2.8kg with battery), simple design with a belt worn around the hips and thighs was created to reduce the wearer's load and to fit different body shapes. Figure 1.4(b) shows the patent of WADSM which has a special mechanism shown as Fig 1.4(c). Because of the special mechanism, the thigh can be lifted easily. Figure 1.4(d) shows the Honda's second walking assist device - Walking Assist Device with Bodyweight Support Assist (WADBSA). It helps support bodyweight to reduce the load on the user's legs while walking, going up and down stairs, and in a semi-crouching position. This could lead to reduced fatigue and less physical exertion. Figure 1.4(e) shows the patent of this device. Although this device is heavier than the first one, it has a very special design. Different from other exoskeleton which are worn on thigh or knee, the device is only worn on shoes as shown in Figure 1.4(e). There exists a place to be seat on and it has force sensors on feet. With the computer and sensors, the device can get specific information to control and drive the motor.

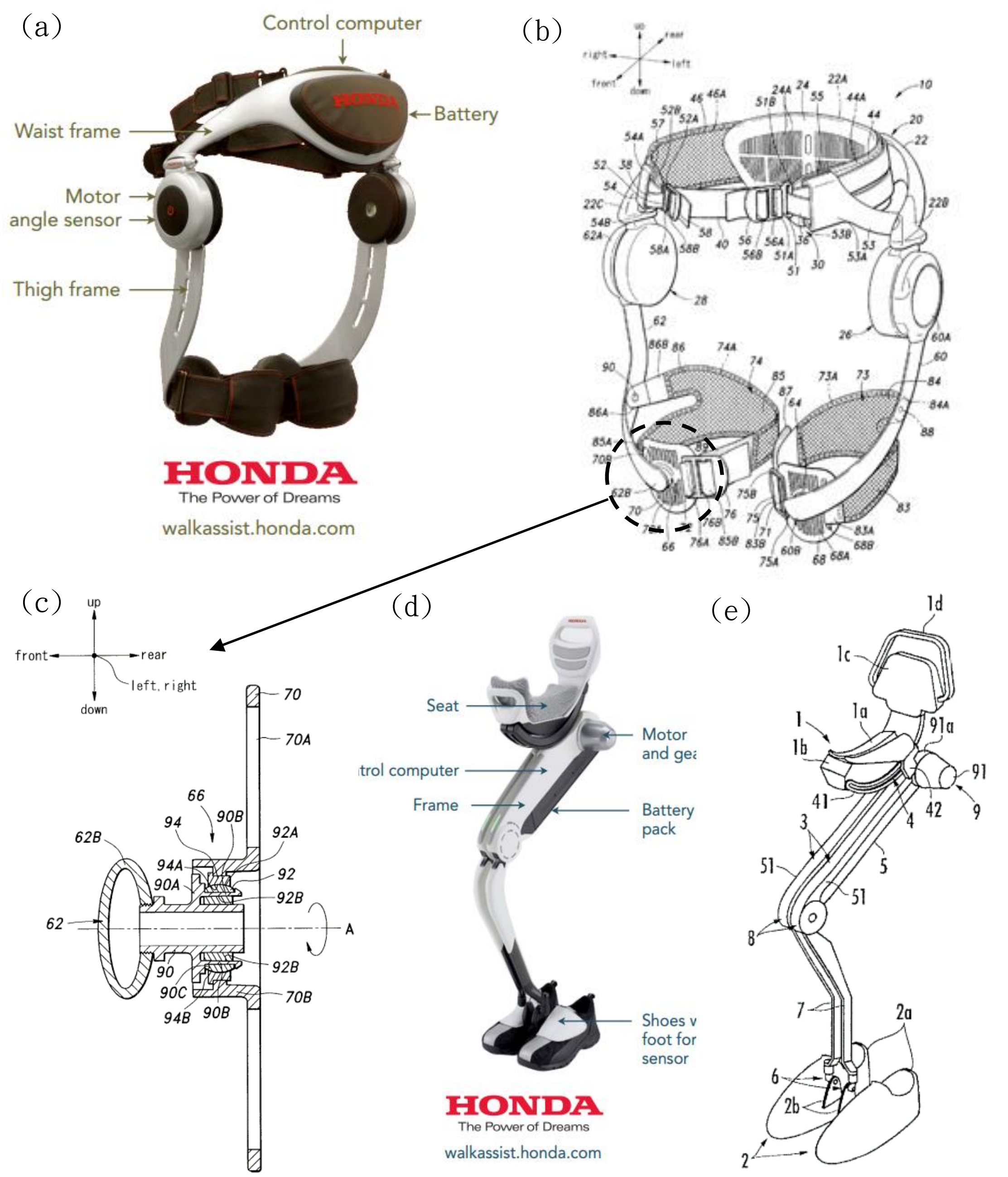
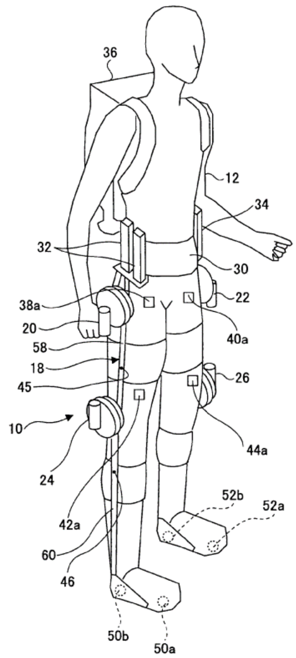


Figure 1.4 HONDA walking assist device (a) Walking Assist Device with Stride Management Assist [17]. (b) Patent of Walking Assist Device with Stride Management Assist [17]. (c) Special mechanism of Walking Assist Device with Stride Management Assist. (d) Walking Assist Device with Bodyweight Support Assist [18]. (e) Patent of Walking Assist Device

Yoshiyuki Sankai and his research team developed the first HAL (Hybrid Assistive Limb) prototype in 1990. The third HAL prototype, developed in the early 2000s, was attached to a computer. Its battery weighed nearly 22 kg and required two helpers to put on, making it very impractical. By contrast, later HAL-5 model weighs only 10 kilograms and wears more conveniently [19]. Figure 1.5(a) shows HAL-5 which is divided into upper and lower limbs. HAL-5 consists of controller/computer, battery, myoelectricity sensors, angle sensors, force sensors, floor reaction force sensors (COP/COG sensors), etc. [20]. As shown in Fig 1.5(b), the patent of HAL-5 lower limbs shows myoelectricity sensors which are marked as 38a, 40a, 42a, and 44a on the thigh in the picture. The force sensor is marked as 45 and floor reaction force sensors are marked as 50a, 50b, 52a, and 52b in the picture. Angle sensors are mounted with power units. With the sensors and power units above, HAL-5 can enhance and upgrade the human capabilities, using sensors to obtain the information from human body and control the device. Actually, most applications of HAL-5 only use lower limb part. And HAL-5 was used in care for the elderly in Japan. Fig 1.5(c) shows the detailed illustrations of HAL-5, number 1 is the computer, number 2 is the controller used to adjust for comfort as shown in Fig 1.5(d), number 3 is the battery, number 4 is the myoelectricity sensor, number 5 is the motor, and number 6 is the reaction force sensor.

(a)

(b)



(c)

(d)

Figure 1.5 Assist device : HAL (a) HAL: Hybrid Assistive Limb [21]. (b) Patent of Hybrid Assistive Limb [22]. (c) Introduction of Hybrid Assistive Limb [21]. (d) Controller of Hybrid Assistive Limb [21].

Finally, we want to introduce the exoskeleton which had gotten FDA-approved. ReWalk^{TM} [23] is a wearable robotic exoskeleton that provides powered hip and knee motion to enable individuals with spinal cord injuries (SCI) to stand upright, walk, turn, and climb and descend stairs. Figure 1.6(a) shows ReWalk^{TM} and its pilot, ReWalk^{TM} was designed for all day use, the battery-powered system features a light, wearable exoskeleton with motors at the hip and knee joints. Combined with the tilt sensor, the ReWalk^{TM} controls movements using subtle changes in his/her center of gravity. Figure 1.6(b) shows the patent of ReWalk^{TM}, mark 23 is the tilt sensor which may be worn on a shoulder strap that holds controller pack (mark 22) to user's torso, and thus senses the degree of tilt of the torso. The tilt sensor may include accelerometers and gyroscopes. The ground force sensors (mark 28) are mounted on each foot places (mark 26). With the sensors above, a forward tilt of the upper body is sensed by the system which initiates the first step, and repeated body shifting generates a sequence of steps which mimics a functional natural gait of legs. Currently, ReWalk^{TM} has already been sold in many countries, and cooperated with hospitals in Taiwan. But if you want to buy an own one, the cost is very high. The above walking assist devices are summarized into Table 1.2.

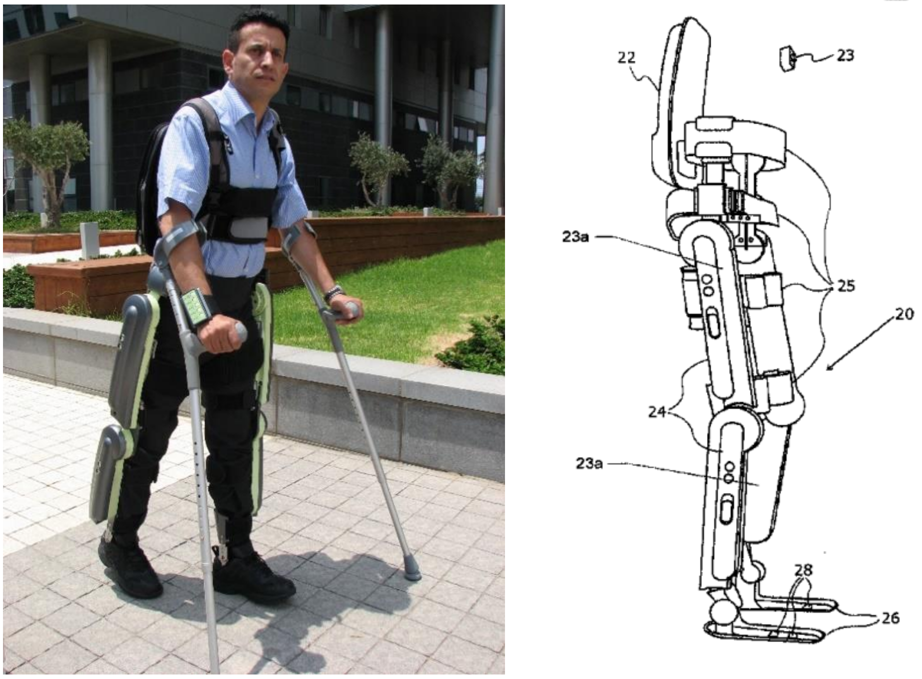


Figure 1.6 REWALK^{TM} (a) REWALK^{TM} and its pilot [23]. (b) Patent of REWALK^{TM} [24].

Table 1-2 Detail of assist device in introduction

Assist devices	Type	Function	Characteristics
ExoHiker™	Enhancement	Carry baggage and walking	Enhanced endurance / reduce the burden
ExoClimber™	Enhancement	Carry baggage and hiking	Enhanced endurance / reduce the burden
UCLC™	Enhancement	Carry baggage and reduce oxygen consumption	Reduce the oxygen consumption
WADSM	Assistance	Assist walking	Lightly
WADBSA	Assistance	Assist walking	Reduce the weight while walking
HAL-5	Assistance/ Enhancement	Assist walking/ Enhancement/ Rehabilitation	Myoelectricity sensors
ReWalk	Assistance	Assist walking/Rehabilitation	FDA-approved

Although the assist devices above provided with multifunction have lots of sensors and excellent control systems, the price of the assist devices are very high. As a result, patients need lots of money to get the device. Since there will be a population explosion in the elderly. If the cost of the assist device is very high, there must be many elders who can’t afford assist devices. In our opinions, we want to design an assist device with safety on walking and prevention of falling. We use simplified mechanism and lower the cost in order to make the elderly be able to have their own walking assist devices so the elderly can take care of themselves.

Chapter 2 Walking assist device design

2.1 Design target

2.1.1 Lower limb joints

In this chapter, we will introduce the details of our design process. Our purpose is to design a walking assist device which can help the elderly who have the ability to walk, but can't walk like a normal person or they need a crutch to help walking. Besides, with specific mechanism, our assist device have another function to help the elderly prevent form falling down. First we start with the human's joint to show how we design the assist device.

We only discuss the hip joint and knee joint because our assist device only contains from hip joint to knee joint. First, we talk about the hip joint. In order to know how the hip joint works, we can get information from International Society of Biomechanics (ISB) who establish the joint coordinate system in order to make the application and interpretation of biomechanical findings easier. First we will introduce pelvic coordinate system and femoral coordinate system before the introduction of hip coordinates since hip coordinates system is related to these two coordinate systems. Figure 2.1 shows that the pelvic coordinate system-XYZ, O is the origin coincident with the right (or left) hip center of rotation. Z-axis is the line parallel to a line connecting the right and left anterior superior iliac spines (ASISs), and pointing to the right. X-axis is the line parallel to a line lying in the plane defined by the two ASISs and the midpoint of the two posterior superior iliac spines (PSISs), orthogonal to the Z-axis, and pointing anteriorly. Y-axis is the line perpendicular to both X-axis and Z-axis points cranially. And then femoral coordinate system-xyz, o is the origin coincident with the right (or left) hip center of rotation, coincident with that of the pelvic coordinate system (O), y is the line joining the midpoint between the medial and lateral femoral epicondyles (FEs) and the origin, and pointing cranially, z is the line perpendicular to the y-axis, lying in the plane defined by the origin and the two FEs, pointing to the right, x is the line perpendicular to both y-axis and z-axis, pointing anteriorly. Before talking about hip joint coordinate system, we need to discuss the motion in cardinal planes first. It can help us understand the rotation of hip joint easier. Figure 2.2 (a) shows the motion in cardinal planes, the vertical sagittal plane divides the body into right and left masses with respect to the midline, running superior to inferior and anterior to posterior. The frontal plane is another vertical plane, but it divides the body in half along the midline into anterior and posterior masses, running superior to inferior and side-to-side. The transverse plane passes through the body horizontally and divides it into superior and inferior masses, passing anterior to posterior and side-to-side. Finally, the hip joint coordinate system – $e_1e_2e_3$, e_1 is the rotation axis of flexion and extension which rotates in sagittal plane as shown in Figure 2.2 (b). Axis e_1 is fixed to the pelvis and coincident with the Z-axis of the pelvic coordinate system, and axis e_3 is the rotation axis of horizontal adduction and abduction (internal / external rotation) as well as the axis fixed to the femur and coincident with the y-axis of the right (or left) femur coordinate system as shown in Figure 2.2 (b). Axis e_2 is the rotation axis of abduction and adduction which rotates in frontal plane as shown in Figure 2.2 (c). Axis e_2 is the floating axis, the common axis perpendicular to e_1 and e_3. With the discussion above, we can know that how hip joint rotates and how can we design the hip joint of our assist device in accordance with human's movement.

And then we want to talk about the knee joint which consists of meniscus, medial collateral ligament, lateral collateral ligament, anterior and posterior cruciate ligament. Knee is one of the largest and most complex joint in the body and knee joint is a hinge joint with slide degree of freedom.

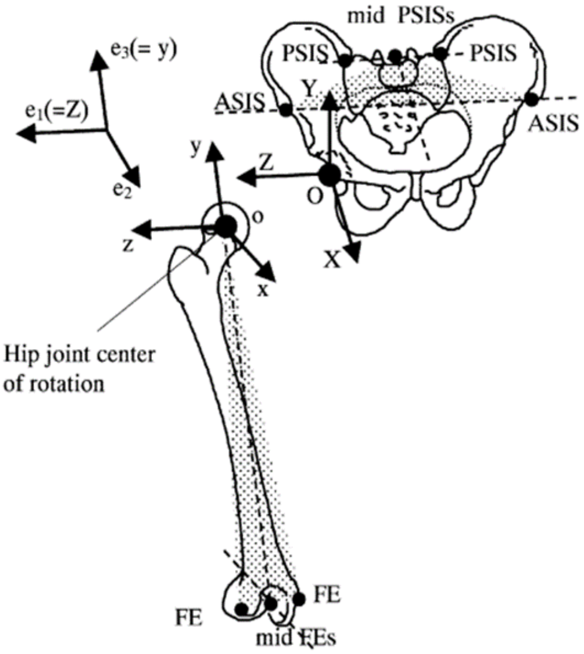


Figure 2.1 Illustration of the pelvic coordinate system (XYZ), femoral coordinate system (xyz), and the joint coordinate system for the right hip joint ($e_1 e_2 e_3$) [25].

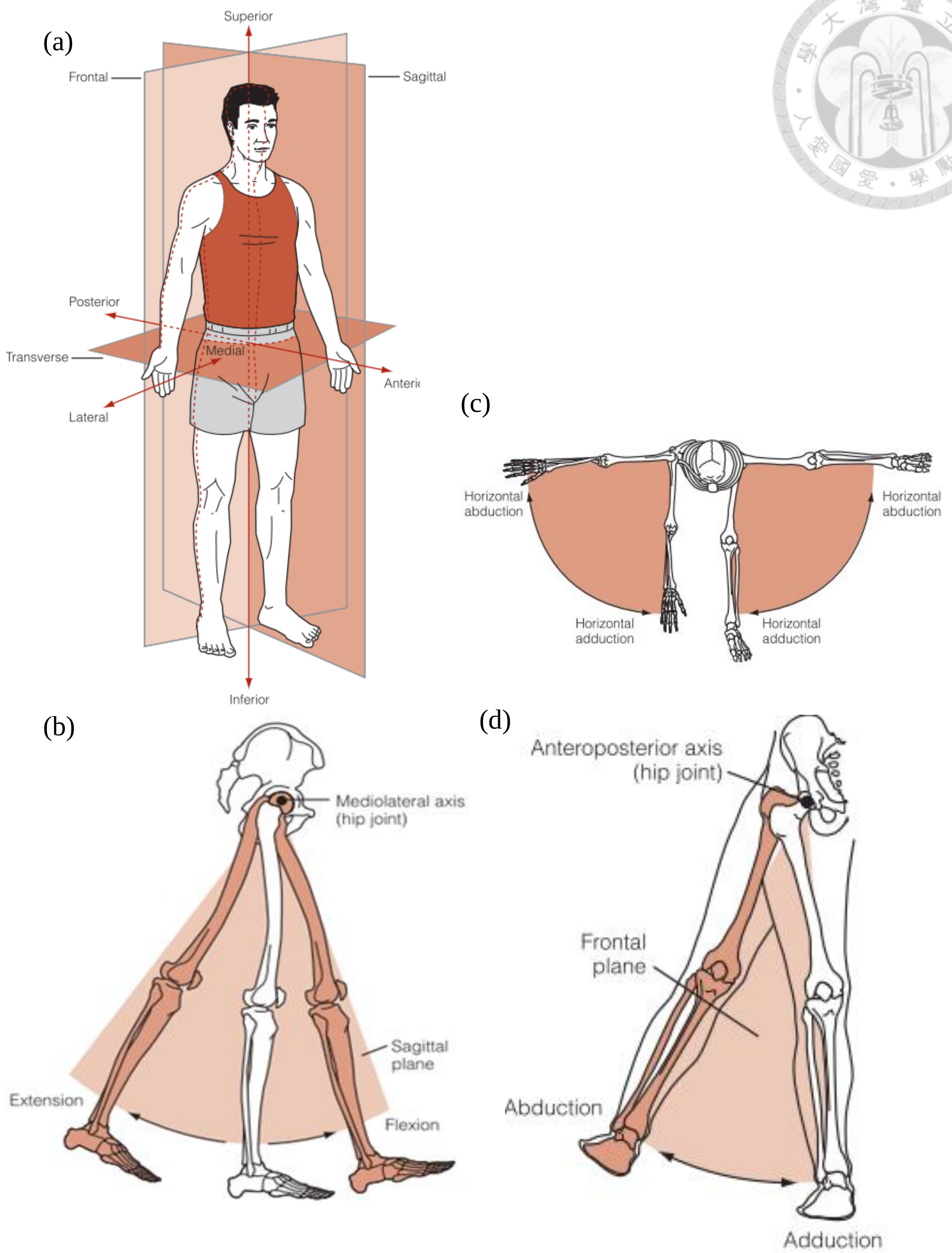


Figure 2.2 Motion of lower limb (a) Cardinal planes of motion [26]. (b) Hip motion of flexion and extension [26]. (c) Hip motion of horizontal adduction and abduction [26]. (d) Hip motion of abduction and adduction [26].

2.1.2 Gait analysis

Gait analysis is the systematic study of animal locomotion, more specifically the study of human motion. Instruments are used to measure the body's information. Gait analysis is used to assess, plan, and treat individuals with conditions affecting their ability to walk. It is also commonly used in biomechanics to help athletes run more efficiently and to identify posture-related or movement-related problems in people with injuries.

Borelli was the first one who researched Biological Kinematics in 1680. Because of the equipment was not fine and the knowledge was not enough at that time, the result of the research was not very well. In 1895, Braune and Fisher started the research of modern biomechanics, they also introduced into the study of life movement the technique of stereometry, of stereophotogrammetry in particular, which today is still the basic experimental technique in biomechanics of human movement. In 1984, Cappozzo introduce a VICON system equipped with three TV cameras and force plate positioned as shown in Figure 2.3(a). Figure 2.3(b) shows the marker configuration and embedded coordinate systems in VICON system. Among them, P_1, P_2, $P_{3,}$ are the markers on pelvis, T_1, T_2, $T_{3,}$ are the markers on thigh, and S_1, S_2, $S_{3,}$ are the markers on calf. In addition to the above-mentioned technical markers, other markers are referred to as anatomical landmarks. In 1990, Kadaba and his research team made reference to a VICON system and they use five TV cameras and force plate positioned as shown in Figure 2.3(c). Figure 2.3(d) also shows the marker configuration and embedded coordinate systems in Kadaba's research. In the Figure 2.3(d), two markers are placed on the right and left ASISs. One other marker is placed on a 10 cm long stick extending from the top of the sacrum (L4-L5) in the spinal plane. Other markers are placed on the following locations of the particular limb under consideration: greater trochanter, rotation axes of the knee joint, shank lateral malleolus, and foot thumb. They find 40 normal healthy testers with no previous history of musculoskeletal problems and measure the rotation of pelvic, hip joint, knee joint and ankle rotation. In this paper, we only discuss the result of hip joint and knee joint, because our design only contains hip joint to knee joint. Figure 2.4 shows the rotation of hip joint and knee joint. In Figure 2.4 (a) to (c), rotations of flexion/extension, adduction/abduction and internal / external rotation of the hip joint are shown respectively. Figure 2.4 (d) to (f) show the rotations of flexion/extension, varus/valgus and internal rotation, external rotation of the knee joint respectively. As the result, we can know the limit in rotation of hip and knee joint. In the rotation of hip joint, maximums of flexion, extension, adduction, abduction, horizontal adduction, and horizontal abduction are about 30°, 10°, 5°, 5°, below 5°, below 5°, respectively. In the rotation of knee joint, maximums of flexion, extension, varus, valgus, internal rotation, and external rotation are 60°,0°,5°,0°, below 5°, below 5°, respectively.

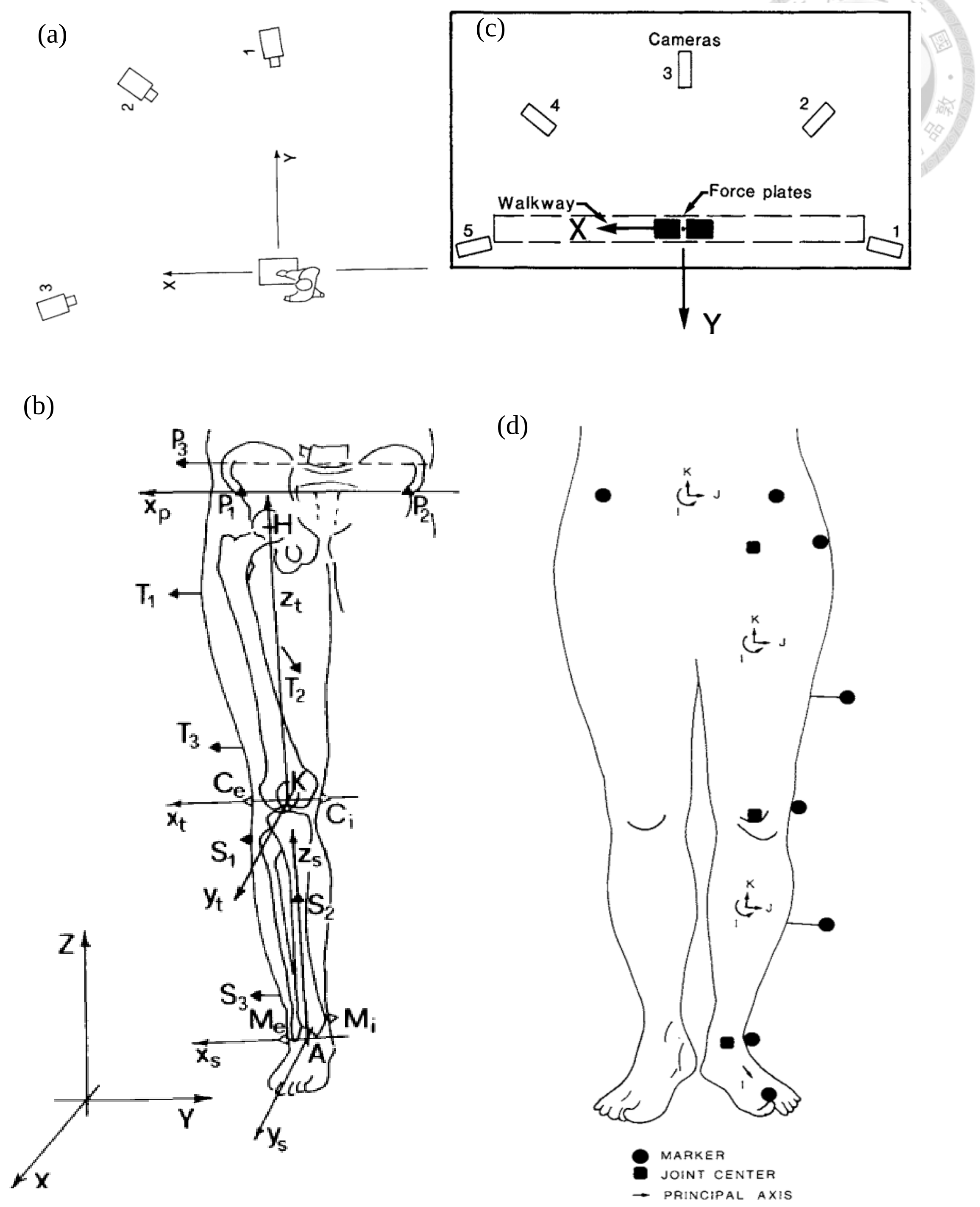


Figure 2.3 Method of gait analysis (a) Stereo metric set up and force plate in Cappazzo research [27]. (b) Marker configuration and embedded coordinate systems in Cappazzo research [27]. (c) Stereo metric set up and force plate in Kadaba's research [28]. (d) Marker configuration and embedded coordinate systems in

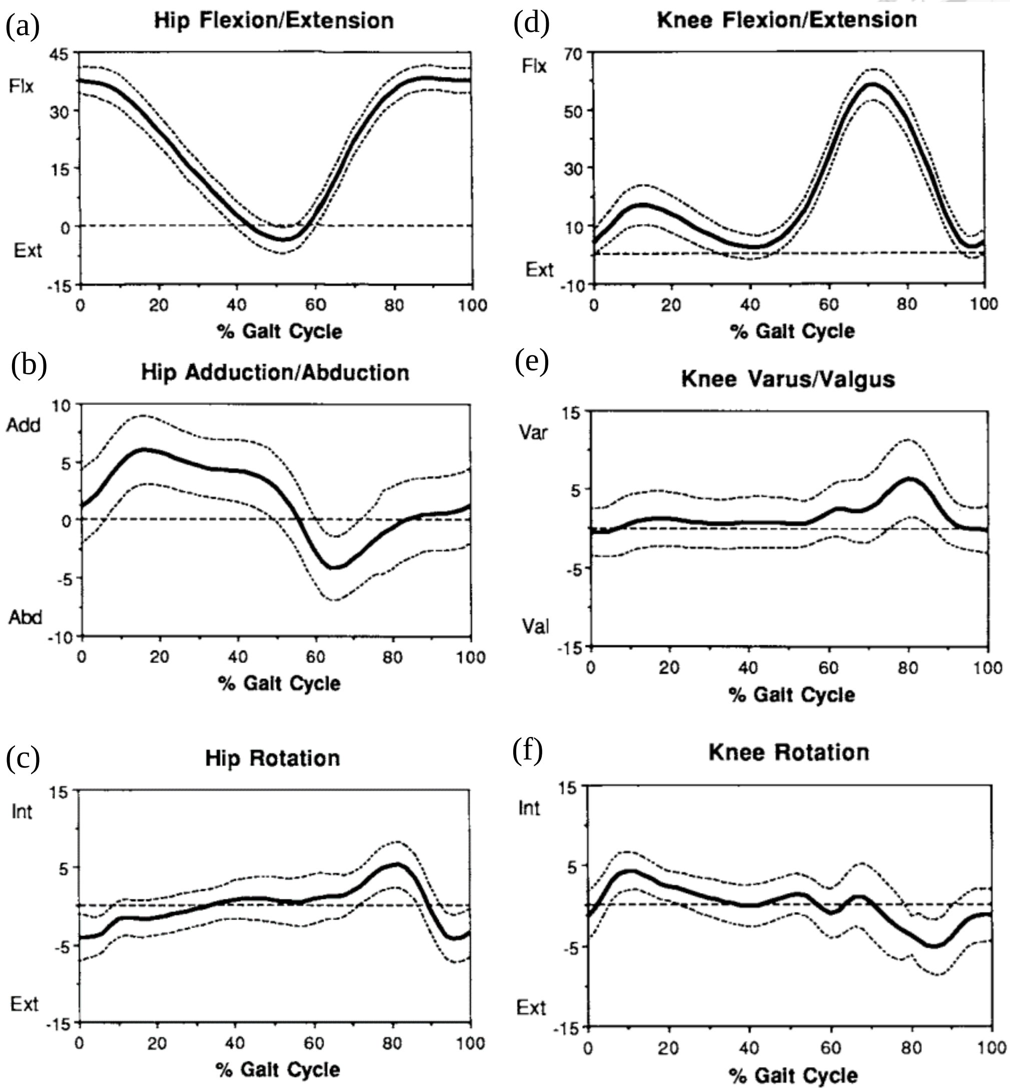


Figure 2.4 Mean (thick line) and one standard deviation (dotted lines) of hip and knee rotation (a) Hip flexion and extension (b) Hip adduction and abduction (c) Hip internal and external rotation (d) Knee flexion and extension (e) Knee adduction and abduction (f) Knee internal and external rotation

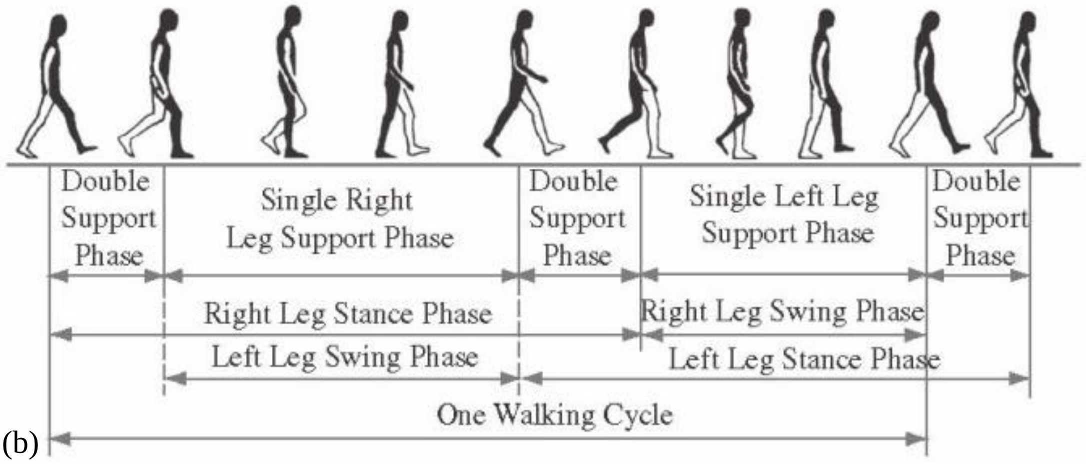
2.1.3 Design target

After discussing the results, we can get a gait cycle, as shown in Figure 2.5(a) in which there are two phases of the gait, double support phase and single support phase. There are also two phases in single support phase, stand phase and swing phase. And then the actual degrees with gait cycle as shown in Figure 2.5(b). Later, we combine the coordinate systems of joint with these gait analysis and design our assist device.

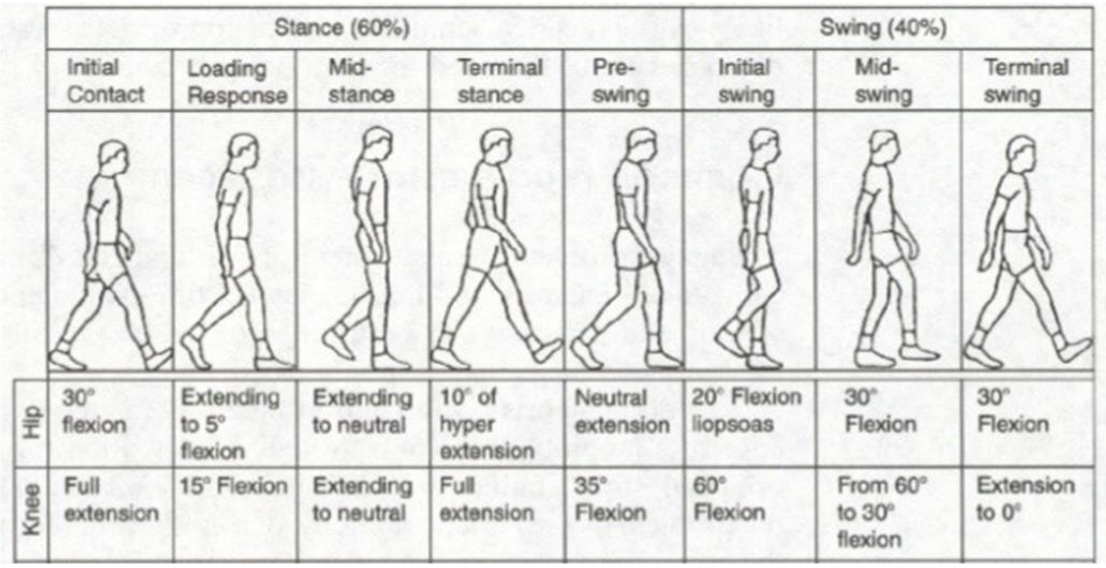
Hip joint is a ball and socket joint with three degrees of freedom, which include rotation of flexion/extension, adduction/abduction and internal / external rotation. Although hip joint is a ball and socket joint, we couldn't design ball and socket joint on the device as hip joint. The assist device is mounted on the human body, and can't be the same as the hip joint in the human body. In order to design a hip joint in our assist device to fit in with the hip joint of human body, we design two axes like the axes of e_1 and e_2 in hip joint coordinate system which defined by ISB above. Axis e_1 is perpendicular to the sagittal plane, and axis e_2 is perpendicular to the frontal plane. We made two axes extend and cross the center of hip joint in the human body and they are perpendicular to each other. According to this condition, the motion of the hip joint on assist device can confirm the motion of human's hip joint as shown in Figure 2.5(c). We ignore the axis e_3 here, because the rotation on axis e_3 is very small. The value which is below 5° is the minimum in all rotation of the hip joint. Also, if we want to design the axis e_3 and confirm the motion of human's hip joint, axis e_3 needs to extend and cross with e_1 and e_2 at the center of hip joint in the human body. With the statement above, it is really hard to design an axis whose rotation is so small on axis e_3 extending to hip joint on assist device, so we don't design it.

As for the knee joint which is a hinge joint with slide degree of freedom, we don’t care about the slide motion and don’t use the motor on knee joint in order to simplify the design. In our design, we provide the assist device to the elderly who have the ability to walk. So we don’t need the motor to control the knee rotation. We use a hinge joint to fit in with the knee rotation, which follows the rotation of human knee. In particularly, we design a sliding groove and slider mechanism for preventing falling on knee. Preventing falling mechanism is combined with the gait analysis and the detail will be described later.

(a)



(b)



(c)

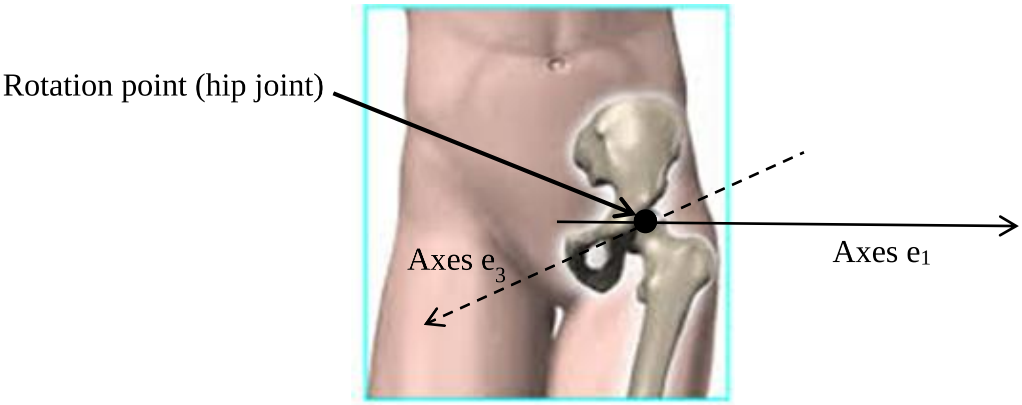


Figure 2.5 Gait cycle (a) Human walking cycle which is divided into single support phase and double support phase [29]. (b) Human's right leg walking cycle with actually degrees [30]. (c) Design purpose on hip joint.

2.2 Device detail

In this section, we will introduce our mechanism of lower limb assist device. Figure 2.6(a) shows the detail of the assist device which contains the component and the three axes on the assist device. There also exists a motor which is mounted on the axis for rotation of flexion and extension. Figure 2.6(b) to (d) show the front view, back view, side view respectively. The middle of the assist device is a square board which has four short shafts and two bigger shafts. The four shafts are for the purpose of worn around the waist with the belt. The bigger shafts are the axes of abduction and adduction, which connect the components of waist. Figure 2.7 shows the components of waist (Figure 2.7(b)) and the components of bar connecting the hip joint and the knee joint. We would call the components of waist and the components of bar connecting the hip joint and the knee joint for waist part and thigh bar respectively. Figure 2.7(b) to (e) show the waist part, top view of waist part, front view of waist part, side view of waist part respectively. There are many holes in Figure 2.7(d), the axis of flexion and extension is mounted on the hole. There exist different sizes of holes in Figure 2.7(e), the bigger hole which represents the axis of motor is surrounded by three small holes. The three small holes are screw holes for the fixation of the motor. And the hole on the top left of the waist part is used to mounted the fixed bar. The waist part is hollowed to make it light. The middle of the hollow place contains a rectangle cylinder which has two holes to be worn on the waist. Figure 2.7(f) to (i) show thigh bar, top view of thigh bar, side view of thigh bar, front view of thigh bar respectively. A hole on the side of the thigh bar as shown in Figure 2.7(h). The hole is a screw hole to mount the rigid shaft coupling on. In Figure 2.7(i), there are four holes. The one on the top is to place the rigid shaft coupling, and the other one on the bottom is the hole for connecting the axis of knee. Two other holes in the middle are the holes mounting the thigh worn part.

Figure 2.8 shows the wear of the thigh and the calf, and we would call them thigh worn part and calf worn part as shown in Figure 2.8(b) and (f) respectively. Figure 2.8 (c), (d), (e), (g), (h), (i) are the top view of thigh worn part, front view of thigh worn part, side view of thigh worn part, top view of calf worn part, front view of calf worn part, side view of calf worn part respectively. The holes on the three-dimensional diagrams above are screw holes for mounting on the device, helping to wear on human body. Finally, Figure 2.9(b) and (f) are the fixed bar for preventing falling and the component on knee, we would call them fixed bar and knee part respectively. Figure 2.9(c) to (d) are the top view of fixed bar, front view of fixed bar, and side view of fixed bar respectively. Figure 2.9(d) shows two holes, the one on the top is mounted on the waist part, and the other one on the bottom is combined with the slider which slides in the sliding groove on the knee part. Figure 2.9 (g) to (i) are the top view of knee part, rear view of knee worn part, and side view of knee part. In the back view of knee part, it has a deep notch. It was difficult to be fabricated, so we separated the knee part into three parts, as shown in Figure 2.9(j). Knee part is separated into two same L-shaped part vise. There are many holes in Figure 2.9(i), the four holes are screw holes for fabricating the knee part, and the other bigger hole is for mounting the axis of knee. There has the rectangle notch in Figure 2.9(i), it provides the sliding groove for preventing falling mechanism. With the slider move up and down in the sliding groove, we can realize the purpose of preventing falling.

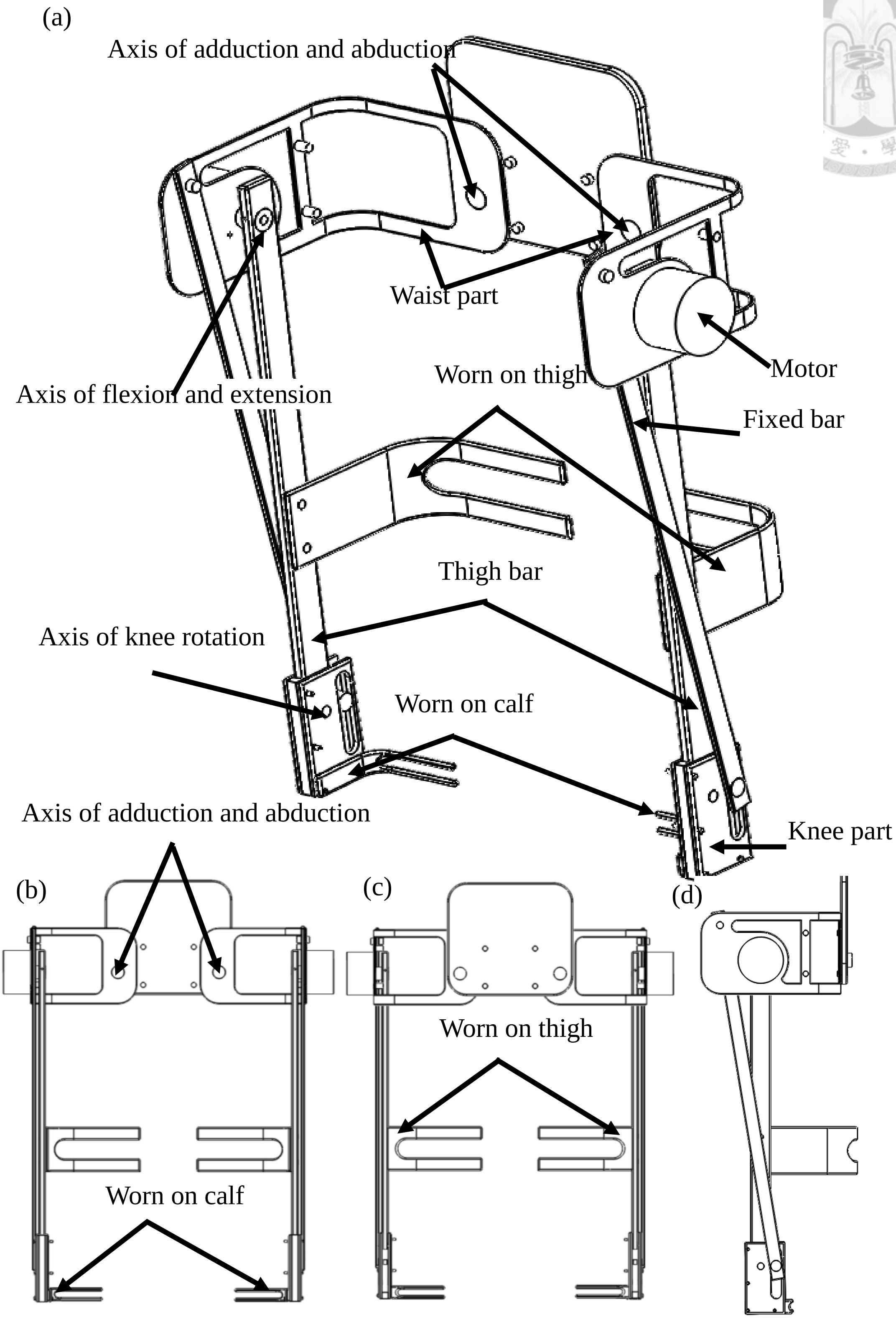


Figure 2.6 Walking assist device (a) Walking assist device (b) Front view of walking assist device. (c) Back view of walking assist device. (d) Side view of walking assist device.

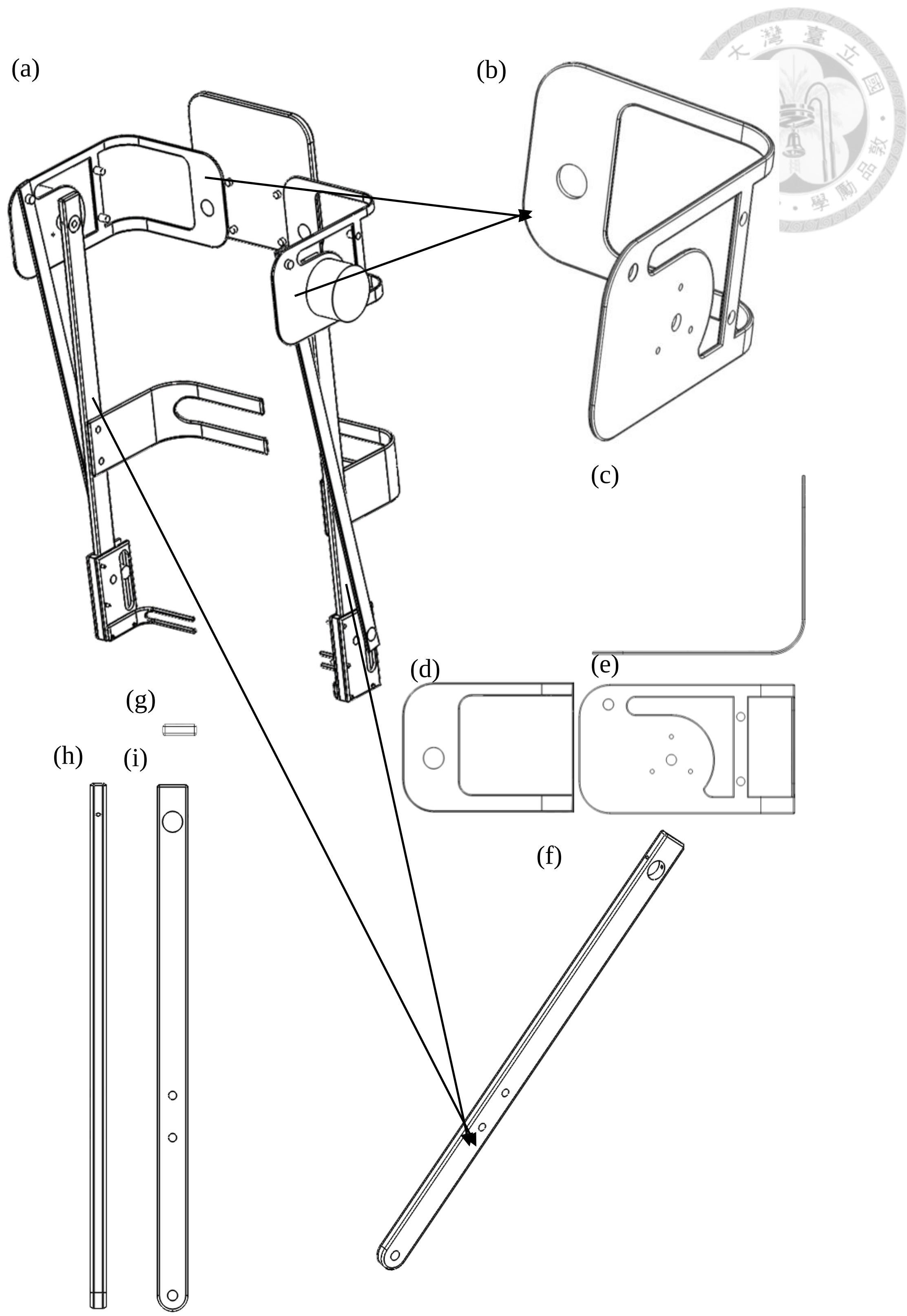


Figure 2.7 Component of assist device (a) Walking assist device. (b) Waist part of walking assist device. (c) Top view of waist part. (d) Front view of waist part. (e) Side view of waist part. (f) Thigh part of walking assist device. (g) Top view of thigh bar. (h) Side view of thigh bar. (i) Front view of thigh bar.

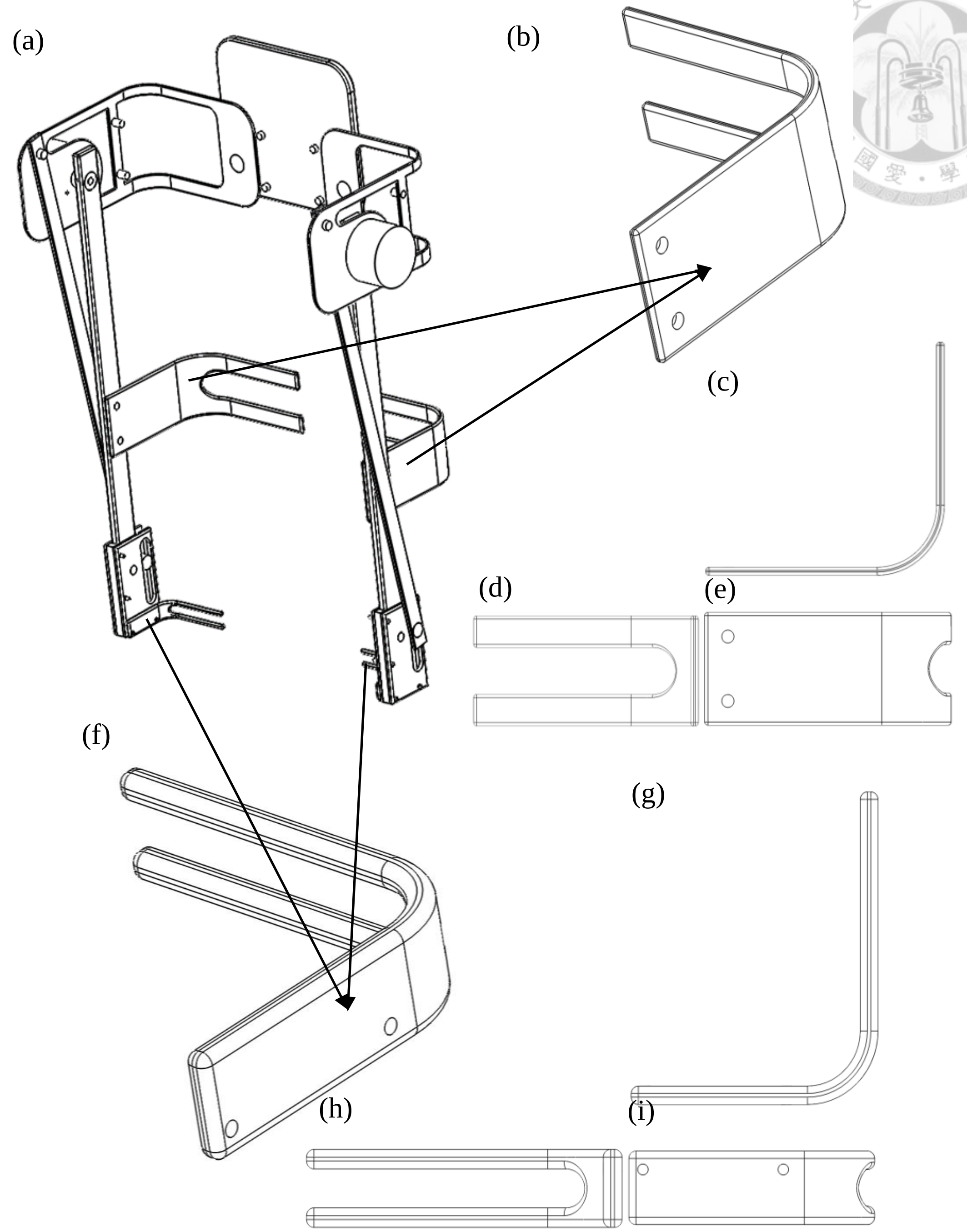


Figure 2.8 Component of assist device (a) Walking assist device. (b) Thigh worn part of walking assist device. (c) Top view of thigh worn part. (d) Front view of thigh worn part. (e) Side view of thigh worn part. (f) Calf worn part of walking assist device. (g) Top view of calf worn part. (h) Front view of calf worn part. (e) Side view of calf worn part.

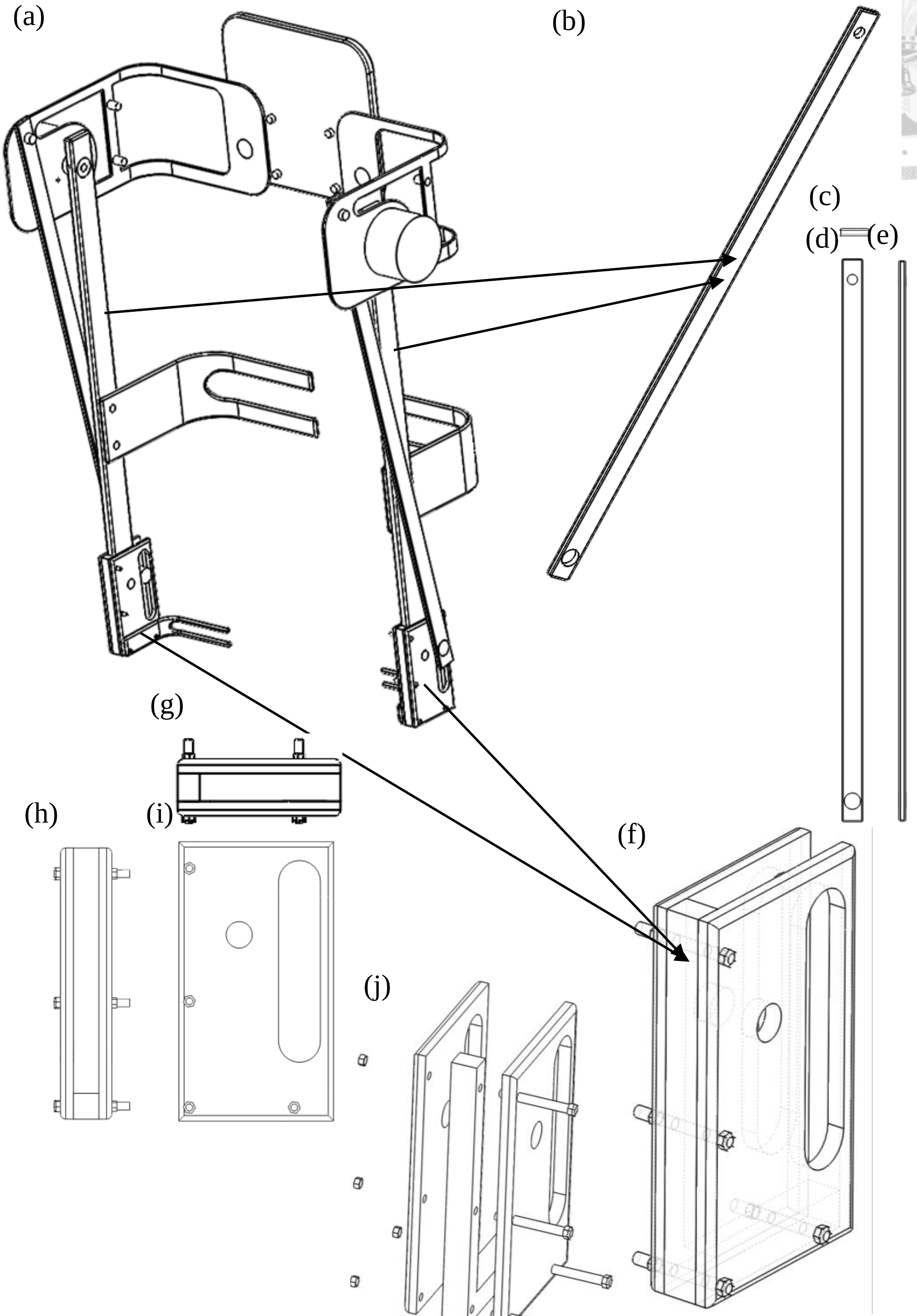


Figure 2.9 Component of assist device (a) Walking assist device. (b) Fixed bar of walking assist device. (c) Top view of fixed bar (d) Front view of fixed bar. (e) Side view of fixed bar. (f) Knee part of walking assist device. (g) Top view of knee part. (h) Back view of knee worn part. (i) Side view of knee part. (j) Exploded view of knee part.

2.3 Preventing falling mechanism

In this section, we would discuss about the operation of preventing falling mechanism. Figure 2.10 shows the side view of walking assist device. The knee part is changeable based on the human gait. The slider inside the sliding groove can differ due to different gaits. From Figure 2.10(b), a simplified path of thigh bar and fixed bar, the knee part which is driven by the thigh bar rotates with the motor axis which is located at the waist part. Since the waist part is worn on a human body, it can be regarded as a fixed end. As for the fixed bar, it is also be seen as a fixed end because of its rotation around the hole of fixed bar on the waist part. We assume that the knee part doesn’t move relative to the thigh bar. That is, there isn’t rotation in the knee joint. The calf forms a straight line with thigh. Based on this assumption, the sliding groove consequently rotates around the motor axis just like the thigh bar. We can observe that the two fixed ends possess their own trajectories while they are rotating. However, the rotation of the fixed bar is driven by the thigh bar, and the lengths of the two bars are different. If the fixed bar rotates the same angle as the thigh bar, the two bars will stuck due to different trajectories. As a result, we add a slider and sliding groove to generate a new degree of freedom, avoiding being stuck. The fixed bar will have a new rotating angle due to the slider. Furthermore, we apply this method to restrict the relative rotation between the thigh and calf for preventing falling mechanism. First, we must know the probable motion caused by the slider to decide a suitable time to fix the thigh bar. The relative rotation between the thigh bar and knee part is expected to be fixed in the instable gait of single support phase, restricting the relative rotation between the thigh and calf of human body. As for the influence caused by the position of slider, we consider the widths of thigh bar and knee part. Since the edge of thigh bar fits in with the sliding groove, the slider can move fluently when the thigh bar is attached to the knee part as shown in Figure 2.10(c). On the other hand, the fixed bar slides up and down as the thigh bar rotates clockwise and counterclockwise respectively. And then with the gait analysis, both right and left legs rotate clockwise after touching the ground when they are under single support phases as shown in Figure 2.4(a) and (b). The slider may therefore slide up based on the above characteristic. Figure 2.10(d) shows the inner structure of the knee part. The height of the slider is equal to the axis of knee part initially. The thigh bar can consequently rotate when the slider slides down, providing a degree of freedom for the rotation of knee joint. And when the slider moves up, the fixed bar will seize the slider to restrict the downward rotation of the thigh bar. Therefore, the relative motion between the thigh and calf is restricted under single support phase. And under swing phase, the thigh bar rotates counterclockwise to make the slider move downward. Thigh again receives a relative rotation with respect to calf. Under above situation, we can provide the walking assist device with both fixation and free capabilities to prevent falling. As for the assumption of the knee part doesn’t move relative to the thigh bar. It don’t influence preventing falling mechanism because when the hip rotation rotate in clockwise, it don’t have any rotation on knee joint. The rotation on knee joint will happen on hip joint to neutral and the hip joint rotate in counter clockwise.

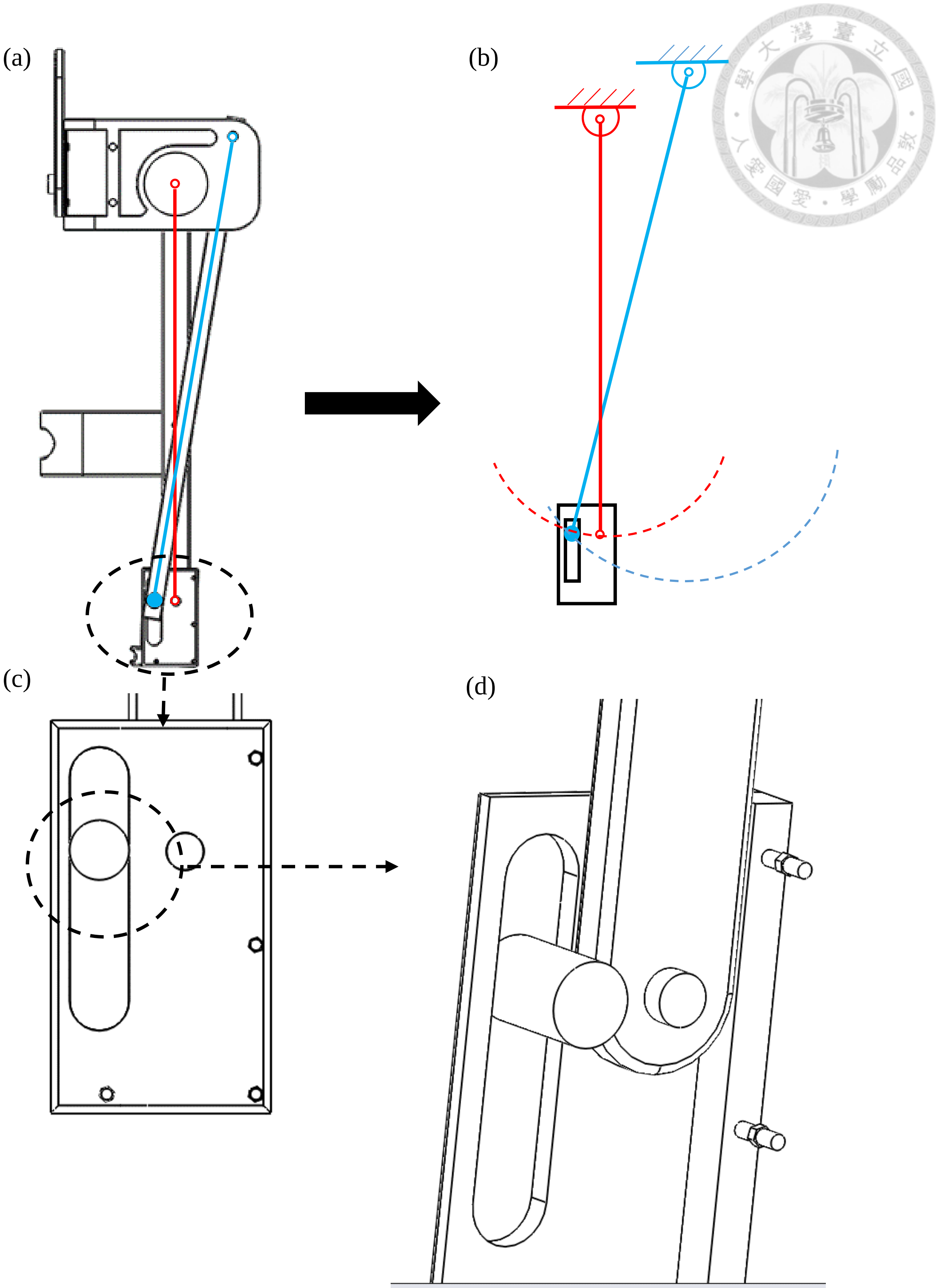


Figure 2.10 Preventing falling mechanism (a) Side view of walking assist device. (b) Path of thigh bar and fixed bar. (c) Side view of knee part (d) Knee anatomy.

2.4 Mechanism model

2.4.1 Mechanism model

We discuss model of our assist device and restriction in preventing falling system. Based on our assumption above, we draw a side view of walking assist device in Figure 2.11. We set the length between hip joint to knee joint as r_1 , the length between the rotation center of knee to the slider center as r_2 , the change length of the slider in the sliding groove as r_3 , and define moving upward as positive, moving downward as negative. The length of the fixed bar is r_4 , the length from motor shaft to the center of the fixed end shaft is r_5 . And $\theta_1,\theta_2,\theta_3,\theta_4,\theta_5$ are the angle between r_1,r_2,r_3,r_4,r_5 and level line. And we set the rotation of hip joint is α and the rotation of knee joint is β , and define counter clockwise is positive and clockwise is negative. According to the vector loop in the figure, we can obtain the equation below

$$\vec{r_1}+\vec{r_2}+\vec{r_3}=\vec{r_4}+\vec{r_5} \tag{2.1}$$

We divide the vector to x and y direction and as shown in equation (2.2) and (2.3).

$$r_1\cos\theta_1+r_2\cos\theta_2+r_3\cos\theta_3=r_4\cos\theta_4+r_5\cos\theta_5 \tag{2.2}$$

$$r_1\sin\theta_1+r_2\sin\theta_2+r_3\sin\theta_3=r_4\sin\theta_4+r_5\sin\theta_5 \tag{2.3}$$

Among of above, r_1,r_2,r_4,r_5 and θ_5 are constant, $\theta_1,\theta_2,\theta_3,\theta_4$ and r_3 are variable. The variable value would change by the rotation of hip joint α and the rotation of knee joint β which have specific value at specific gait. Because of the specific degrees within the gait, we can know $r_1, r_2, \theta_1, \theta_2, \theta_3$ above, r_1 is defined by the length of thigh (hip joint to knee joint). r_2 , which is due to the length of knee joint to the center of the sliding groove, and r_2 is perpendicular to r_3 . θ_1 is depended on the rotation α . θ_2 and θ_3 are depended on α and β whose value can be known with the gait cycle. The degree can show in equation (2.4) to (2.6)

$$\theta_1 = 270^\circ + \alpha \tag{2.4}$$

$$\theta_2 = 180^\circ + \alpha + \beta \tag{2.5}$$

$$\theta_3 = 270^\circ + \alpha + \beta \tag{2.6}$$

And then, with equation (2.2) and (2.3), we have two equations but r_3 $r_4, r_5, \theta_4, \theta_5$ are unknowns. We don't have enough equations to solve the unknowns. With the different gait, we can get two more equations. But r_3 and θ_4 is variable, we get two more unknowns with the two more equations. The unknown still can't be solve, so we need to discuss the value of r_3 for reducing unknown and solving value for the mechanism. r_3 is an important value on preventing falling which is the displacement of the slider in preventing falling mechanism and changed by the gait. We have talked about the preventing falling mechanism in section 2.3. Although r_3 is variable value with the gait changing, we set specific value on the specific gait for preventing falling mechanism. First, we set the height of the slider center same as the knee joint when $\alpha = 0$ as shown in Figure 2.11 (a) and (b). It benefits for preventing falling mechanism in which the slider moves upward when α rotates clockwise and moves downward when α rotates counterclockwise. So we can know that when $\alpha = 0, r_3 = 0$ and equation (2.2) and (2.3) can be rewritten as

$$r_2 = r_4 \cos\theta_{4(\alpha=0)} + r_5 \cos\theta_5 \tag{2.7}$$

$$r_1 = r_4 \sin\theta_{4(\alpha=0)} + r_5 \sin\theta_5 \tag{2.8}$$

In equation (2.4) and (2.5), there are two known: r_1 and r_2 four unknowns: $r_4, r_5, \theta_{4(\alpha=0)}, \theta_5$. Comparing with the equations and unknowns above, we only emerge one more unknown $\theta_{4(\alpha=0)}$, but we have two more equation. We reduce unknown with setting the specific r_3. And then, we set another specific r_3 on other specific gait. Figure 2.11 (c) and (d) show the thigh bar rotating to 10° clockwise, this situation is on stand phase with single support as Figure 2.5(a) showing which is the situation need to fix the relative rotation between thigh and calf we discuss in section 2.3. So we choice a specific r_3 at this gait and the equation (2.2) and (2.3) can be rewritten as

$$r_1 \cos\theta_1 + r_2 \cos\theta_2 + r_{3(\alpha=10^\circ c.w)} \cos\theta_3 = r_4 \cos\theta_{4(\alpha=10^\circ c.w)} + r_5 \cos\theta_5 \tag{2.9}$$

$$r_1 \sin\theta_1 + r_2 \sin\theta_2 + r_{3(\alpha=10^\circ c.w)} \sin\theta_3 = r_4 \sin\theta_{4(\alpha=10^\circ c.w)} + r_5 \sin\theta_5 \tag{2.10}$$

Combing with equation (2.4) to (2.6) the equation can rewritten as

$$r_1 \sin\alpha - r_2 \cos(\alpha+\beta) + r_{3(\alpha=10^\circ c.w)} \sin(\alpha+\beta) = r_4 \cos\theta_{4(\alpha=10^\circ c.w)} + r_5 \cos\theta_5 \tag{2.11}$$

$$-r_1\cos\alpha - r_2\sin(\alpha+\beta) - r_{3(\alpha=10^\circ c.w)}\cos(\alpha+\beta) = r_4\sin\theta_{4(\alpha=10^\circ c.w)} + r_5\sin\theta_5 \qquad (2.12)$$

In equation (2.11) and (2.12), we get two more equations and one more unknown $\theta_{4(\alpha=10^\circ c.w)}$. Combining with equation (2.7) and (2.8) we have four equations and five unknowns. If we set another specific r_3, we can have six equations and six unknowns. But both specific r_3 in equation (2.7) and (2.11) are fit with the preventing falling system on specific gait. We don't have one more restriction to solve the problem. We can set any value of r_3 for solving the unknown. It means we have any value of $r_4, r_5, \theta_4, \theta_5$. But with the preventing falling system, we have the restriction $\alpha = 0, r_3 = 0$, $\alpha > 0, r_3 < 0$ and $\alpha < 0, r_3 > 0$. And others restriction are set with our discussion above. Combining the restriction, we can set the value of r_3 in the range of the restriction. And we have six equations and six unknowns, then we can get actually value. So we choice the suitable value r_3 when α is in counter clockwise. The value would fit with the restriction we would discuss in next section.

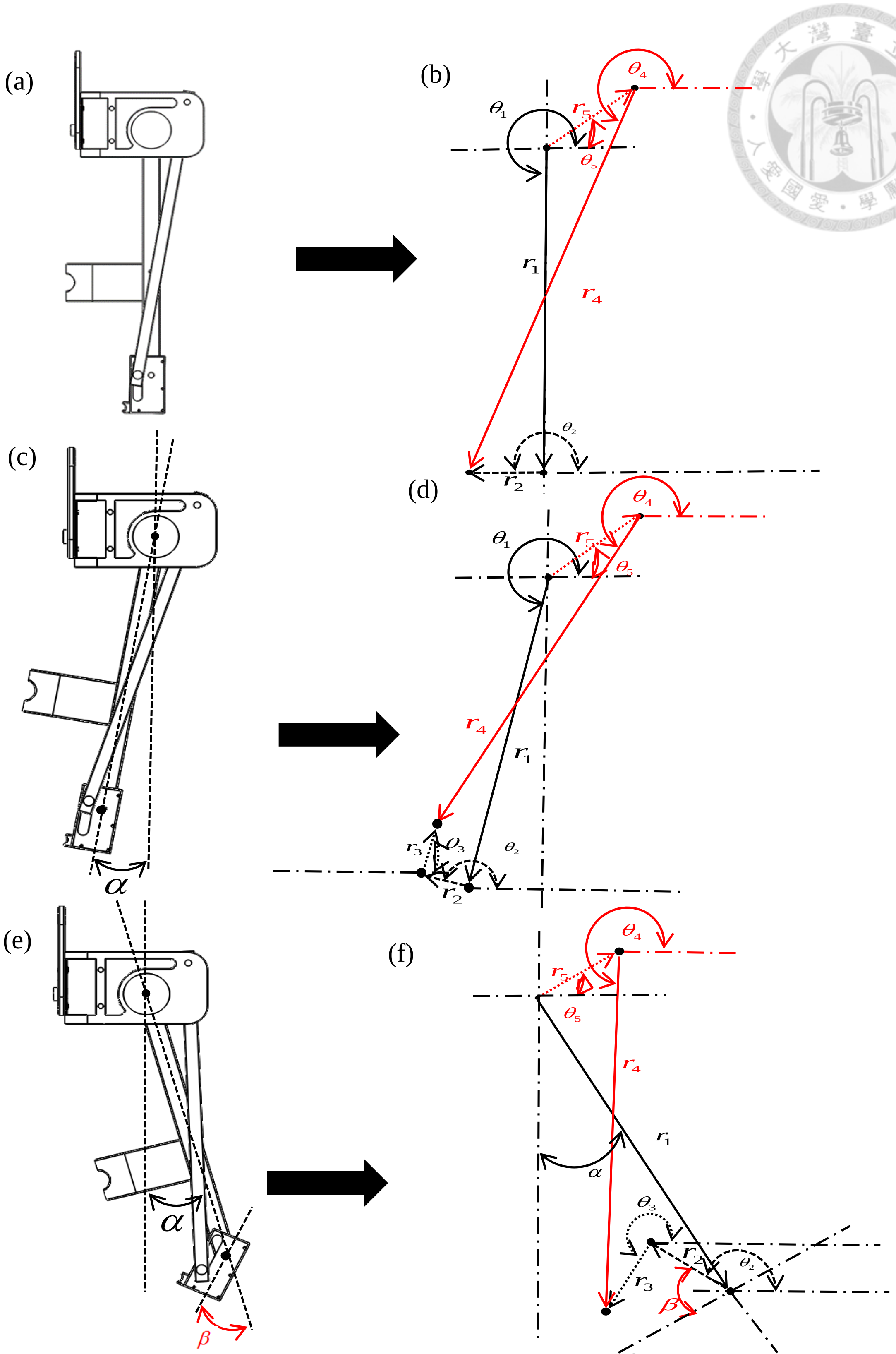


Figure 2.11 Side view of walking assist device (a) In $\alpha = 0$ degree. (b) Geometry in $\alpha = 0$ degree. (c) In clockwise α degrees. (d) Geometry of α clockwise degree. (e) In α counter clockwise degree and β clockwise. (f) Geometry of α counter clockwise degree and β clockwise

2.4.2 Dimension and restriction of the assist device

Next, we need to calculate the actual size of the assist device to manufacture the first prototype. Finding the problem and make experiments on first prototype, and modify it on better size. First, we consider the restrictions in every components depended on the connection or method of fabrication, and then we find the restrictions from human body. Finally we will discuss the restriction in preventing falling mechanism. Combing these restrictions mentioned in this section and the model above, we can make a new assist device directly if we need to make some changes on the sizes of any components.

First we would discuss the restriction on waist. Based on the above discussion of the hip joint, we wanted to make axes (flexion/extension and adduction/abduction) cross at the center of hip joint in human body. Because of that, we need to make two axes at the same height on the waist part, fitting in with the human's hip joint. And then the motor and the fix bar can be mounted on the waist part. The distance of both center needs to be bigger than the motor's radius so that the fixed bar and motor would not collide. The height of the waist part depends on the distance of motor center and the fixed bar. The height of the waist part also depends on the size of the motor. We use sheet-metal to make our waist part, so the thickness is restricted by sheet-metal fabrication. The restriction of the human body on waist part is the length to which need to fit in with the size of human's waist.

As for the thigh bar, the width depends on the knee part for preventing falling mechanism. The thigh bar is the thickest component in our device, because the thigh bar would support some weight of human's thigh. And the length of thigh bar depends on the length between human's hip joint to knee joint. To make sure that the thigh bar covers human's thigh tightly, the thigh bar is worn on the middle segment of human's thigh since that the diameter of human's thigh raises among the height.

The length of the knee part is fitted in with the height of human's knee and the sliding groove. Because of preventing falling mechanism, the width of the knee part depends on the width of thigh bar and the sliding groove. The thickness of the knee part is based on the thigh bar. And then the wear on the knee is set at the bottom of the knee part in order to make relative rotation between the thigh bar and knee part easier.

Then, we add the restrictions of the components and combine with the equations (2.7) to (2.12). First, we talk about the waist part. Because of the sheet-metal fabrication, we set the thickness of waist part as 3mm for fabricating conveniently. The waist part need to be surrounded the wearer's waist. As the result, we set 165mm and 210mm as the lengths of the front and side of the waist part respectively. Based on sheet-metal fabrication, the waist part and thus the heights of front view and side view of the waist part are same. The size of height depends on the size of motor. Figure 2.12 shows the detail of Maxon motor EC-60 flat. Since we know the size of the motor, we can get the restriction information that r_5 is smaller than 34mm.

Next, we talk about the thigh bar, we choose 10mm to be the thickness of thigh bar, and then we know that the rigid shaft coupling have 20mm in diameter. We can know our width or hip bar need to be bigger than 10mm. Finally 450 mm is selected to be the length of the thigh bar based on the wearer's thigh.

As for the size of knee part, the wear portion is set at the bottom of the knee part to let the wear more convenient, so that the length of the knee part can be decided by the wearer's knee height. And the width of the knee part depends on the width of thigh bar and the sliding groove in the preventing falling mechanism. In the other hand, we separate the knee part into three parts for convenience. The thicknesses of the three piece parts are 5mm, 10mm, 5mm. The knee part can be assembled via the three pieces. Finally, we talk about the size of the sliding groove. We choose 16mm as the width of the sliding groove since that the slider is a cylinder whose diameter is 16mm. And the length of the sliding groove depends on the path of the slider. The path is chosen based on the limit of the rotation of hip joint (clockwise 10° to counterclockwise 30°).

The length of the fixed bar is r_5 which depends on the preventing falling mechanism. The width of the fixed bar is chosen via the diameter of the fixed cylinder. And then we combine all the restrictions mentioned above to get some restriction terms.

$$r_5 > 34(mm) \tag{2.13}$$

$$r_1 = 450(mm) \tag{2.14}$$

$$r_2 > 10(mm) \tag{2.15}$$

$$\alpha = 0, r_3 = 0 \tag{2.16}$$

$$\alpha < 0, r_3 > 0 \tag{2.17}$$

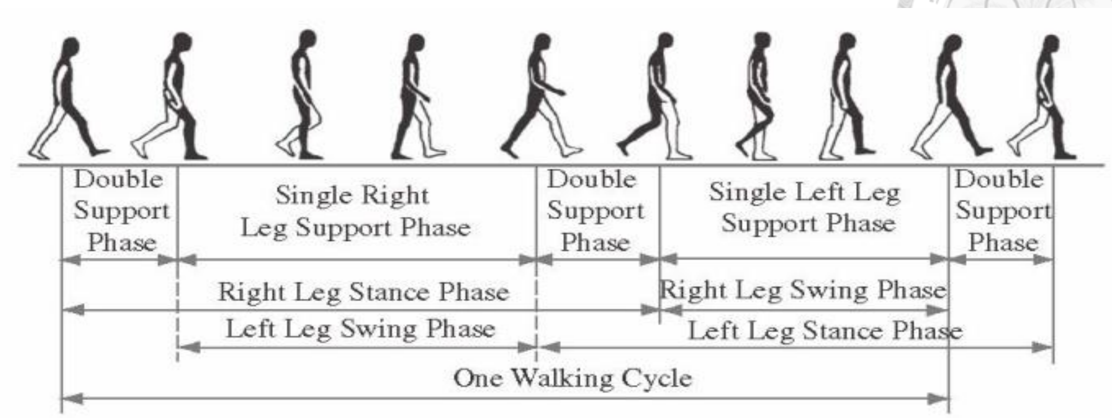
$$\alpha > 0, r_3 < 0 \tag{2.18}$$

At last we need to find the value of r_3, its length which influences the length of sliding groove. Actually, if the height of knee part is constant, the length of the sliding groove is as longer as better. That is, we can reduce the weight of knee part, making the assist device lighter. When the upward value of r_3 is bigger, the preventing falling mechanism would be better. However, the bigger upward value of r_3 would cause the bigger downward value of r_3. It means there must need a bigger knee part to fit the sliding groove. As a result, we set the maximum upward value 20 mm for preventing falling system, which means $r_3 = 20mm$ in equation (2.11) and (2.12) for the preventing falling system. Then, we need choice a value of r_3 when α is counter clockwise and β is clockwise. First, we want to talk about fall-preventing mechanism when $\alpha < 0, r_3 > 0$. We discuss equation (2.17) and combine with equation (2.2) and (2.3). Here r_1, r_2, r_3, r_4, r_5 are the length, must be positive in equation (2.2). Combining with Figure 2.11 (d), x component in equation (2.2) of left side is depended on r_1, r_2, r_3 and r_1 is the biggest within input lengths (r_1, r_2, r_3). It also show in Figure 2.11 (d). So the left side in equation (2.2) is negative and the equation has r_4, r_5 at the right side. r_4 is bigger than r_1 because of fall-preventing mechanism. If r_4 is smaller than r_1, the value of r_3 would not have obvious value and the slider can't work in preventing falling we discuss in section 2.3. The right side of equation (2.2) also need to be negative, but r_4 is bigger than r_1, r_5 need to be positive to fit the equation. Combing with all situation above, r_5 and θ_5 need to fit with the equation. We can know that $\theta_5 \in I, IV$ and it has similar situation on equation (2.3) with equation (2.18) when $\alpha > 0, r_3 < 0$ in Figure 2.11 (f). We can also get $\theta_5 \in I, II$, so θ_5 is in the first quadrant. Combing all discussion above, we need to choose a value of r_3 in another gait. Here we choose the maximum downward value on the gait when hip joint rotates 20° counter clockwise and knee joint rotates 60°. Then we have six equations and $r_4, r_5, \theta_{4(\alpha=0)}, \theta_{4(\alpha<0)}, \theta_{4(\alpha>0)}, \theta_5$ six unknown. We can get the actual size of our mechanism. We would choose length size as integer for fabricating conveniently.

Chapter 3 Simulation of device

In this chapter, we want to use some simple methods to prove the efficacy of our mechanism. First, we simulate our walking assist device with human's gait in CAD. In CAD program, we set our assist device to fit in with the degrees in gait analysis as shown in Figure 3.1. Figure 3.1 shows the gait cycle and the simulation. In the first picture, the gait is at double support phase and the right hip joint rotates 30° counterclockwise. As for the hip joint in left leg, it rotates 10° clockwise and rotation doesn't happen in knee joint. The second picture shows the start of the single support phase. Right leg is at single support phase with a rotation 20° counterclockwise with respect to the hip joint. And the rotation of knee is 20° clockwise. The left leg at this phase is at the swing phase with rotation of hip joint and knee joint for 10° counterclockwise and 35° clockwise. And then the third gait, after the right leg contacts the ground, the center of gravity on the human tilts forward and the rotation of right hip joint goes back to the neutral. So it doesn't have any rotation in the right leg at this phase. At this phase, the rotations of hip joint and knee joint on the left leg are 10° counterclockwise and 50° clockwise. Finally, the fourth gait which are similar to the first gait with smaller degrees and the position of the right leg is exchanged with the left leg. The right leg is at the stand phase and left leg is at the swing phase in first four pictures, being opposite to the last four pictures. And the degrees are 10° and 20° on right and left legs in the fourth picture respectively. As a result, we don't depict the last four pictures in detail. We can complete the whole comparison between gait cycle and our pictures of CAD. The pictures of CAD can be suitable for the gait. According to the discussion above, the mechanism can be realized on actual walking.

(a)



(b)

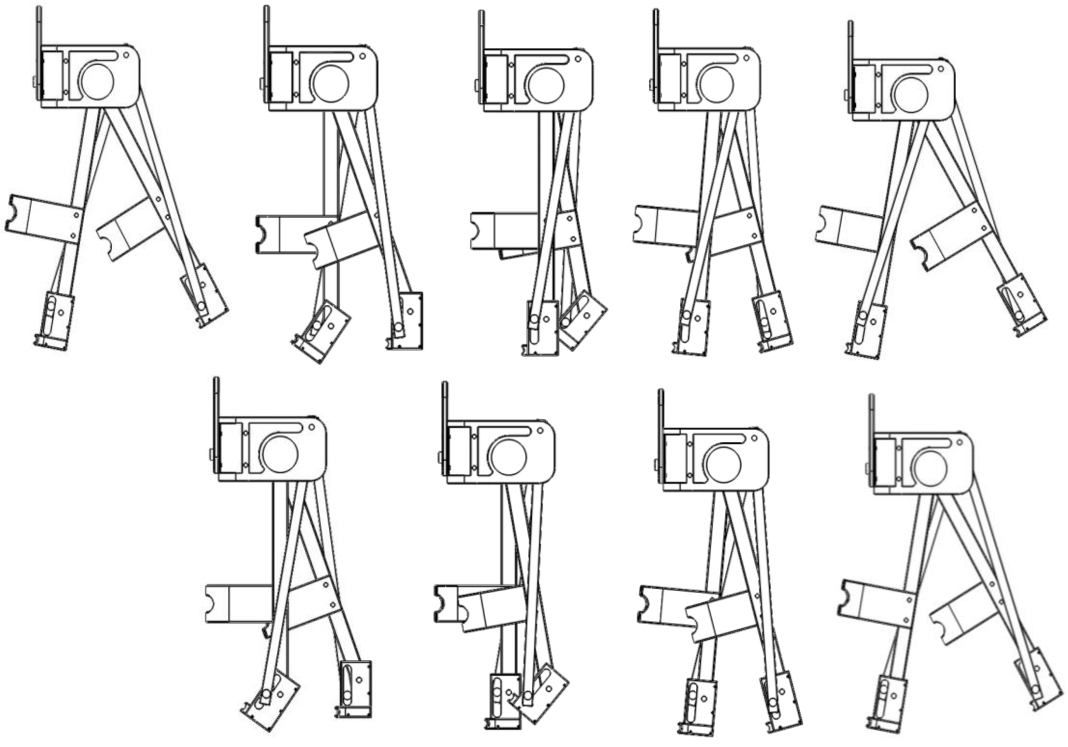


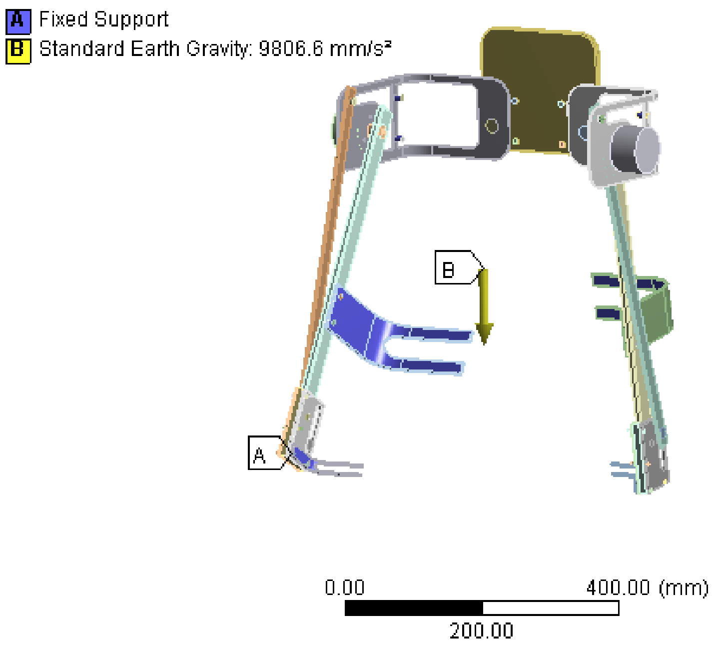
Figure 3.1 Simulation of assist device (a) The gait cycle of human. (b) The gait cycle of the walking assist device.

Next, we use finite element method to make sure our mechanism whether it has problems on its structure. According to the result above, the right leg and left leg exchange with each other and the fourth picture is similar to the first picture. Consequently, we only analyze the limit of mechanism in the first three pictures.

Before setting the boundary condition, we want to discuss the forces applied on the structures. Our purpose is to enhance the elderly's walking ability so that they don't need to use the crutches any more. With the aid of our mechanism, they can take care of themselves. Since our mechanism doesn't cover the ankle and foot, the walking assist device doesn't support the weight of human. That is, the weight of human is supported by themselves on both legs and it means that we can deem the assist is mounted on human body. However, when the gait is at the single support phase, the center of gravity change, so the weight of leg on swing phase may change the center of gravity. We are not sure how much weight is applied on the device during the center of gravity changing. Finally, the weight of the swing leg may focus on the standing leg. So we set two conditions to simulate the mechanism at the single support phase. First one is no weight applied on the device which only supports the weight of itself. Second one is whole human's weight applied on the device. As the assumptions above if the device can support the whole weight, we can claim that our walking assist device is strong enough for all the conditions.

Figure 3.2(a) shows our boundary conditions at double support phase. We make the place where the device is worn on the human body be fixed support, including the waist, calf, and thigh. We don't set the weight of human body here due to the double support phase. In Figure 3.2(b), it is the result of the first gait in Figure 3.1. We can observe that the maximum stress whose value is 3.6 MPa occurs at the place where the fixed bar mounted on.

(a)



(b)

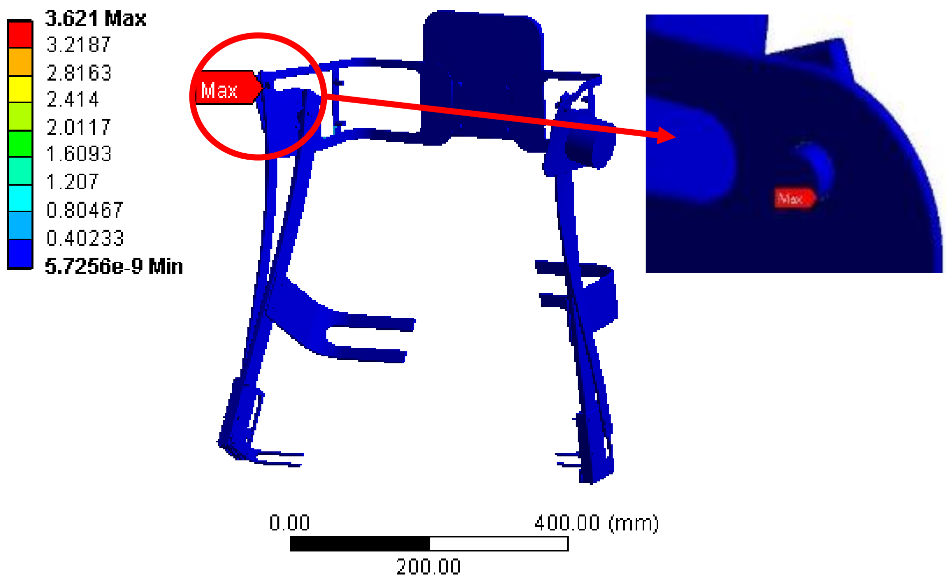


Figure 3.2 Boundary condition and result of first gait (a) The boundary conditions at double support phase. (b) The result and the place where the maximum stress occurs at double support phase.

Before talking about the gait at the single support phase, we need to discuss the weight of human body segments first. We need to know how much weight should be applied on the device. Table 3.1 shows the normalized mass of human body segments. The weight of our wearer is about 70 kg in total, so they are 7 kg to both thighs and 3 kg to both calves. These values are consequently used in our boundary conditions.

Table 3-1 Normalized mass of body segments (standard human) [31]

Segment	Segment mass/ Total body mass
Hand	0.006
Forearm	0.016
Upper arm	0.028
Foot	0.015
Lower leg(calf)	0.065
Upper leg	0.100
Head and neck	0.081
Trunk	0.497

And then we discuss the simulation at the single stand phase of the gait cycle in Figure 3.1. The right leg is at the stand phase and the left leg is at the swing phase at the second gait in Figure 3.1. Therefore, the boundary conditions in second gait are set to be the same as the two boundary conditions discussed above. In the first case, we don't apply the weight on the device as shown in the Figure 3.3(a). And in the second case, we apply the weight on. The weights of both thighs and both calves are applied on specific parts in our device as shown in Figure 3.3(b). In Figure 3.3(b), the whole weight of thigh (70N) is separated into four parts since the weight needs to be averagely distributed in the device. The same situation can be set on the calf portion in the third gait. And then Figure 3.4(a) and Figure 3.4(b) show the results of the first boundary condition and the second boundary condition respectively. We can observe that the maximum stress whose values are 16.6 MPa and 91 MPa for the two cases occur at the places where the fixed bars are mounted on.

Finally we talk about the simulation of the third gait in Figure 3.1. In this gait, we set the boundary condition the same as the second pictures and the condition can show in Figure 3.5(a) and (c). And then we can get the results with or without weight applied respectively as shown in Figure 3.5(b) and (d). In Figure 3.5(b) and (d), the maximum stresses whose values are 9.2 MPa and 92 MPa occur at the places where the fixed bars are mounted on.

In result, we can know that the yield strength of the material must exceed 92 MPa so as to be strong enough no matter how much weight is applied on the device.

(a)

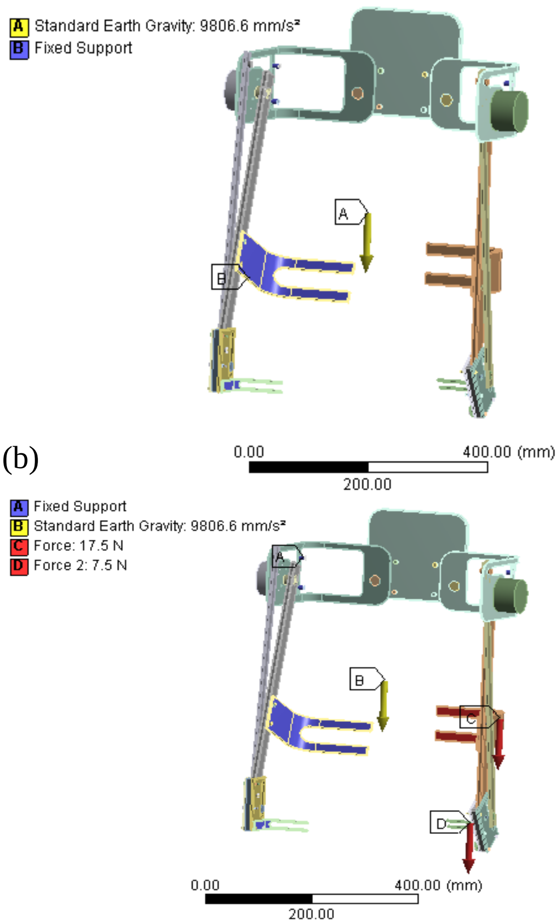
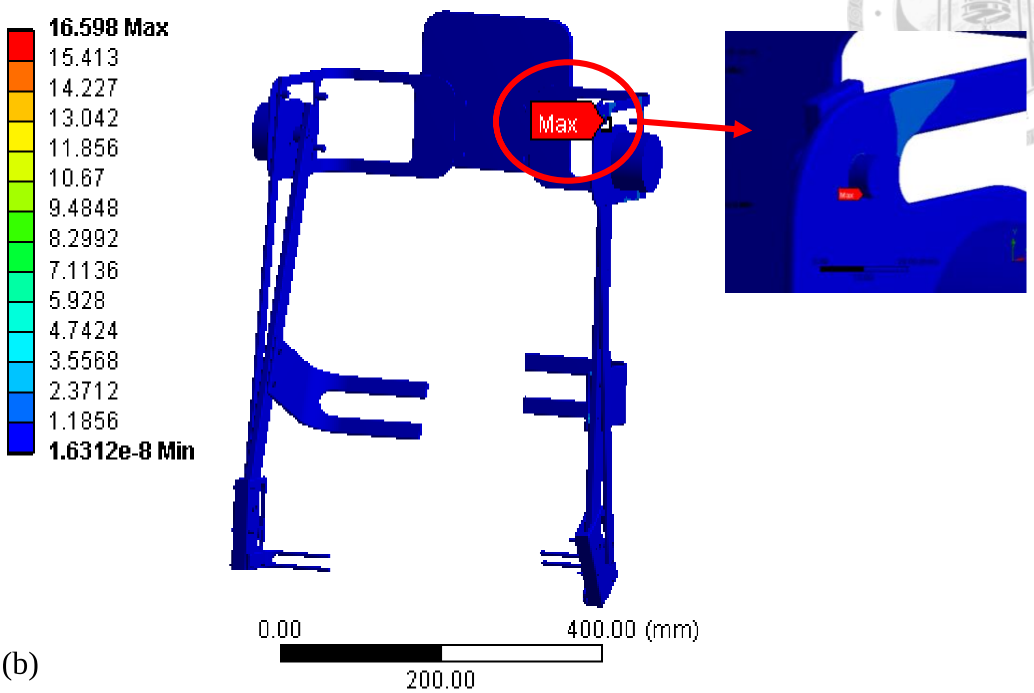


Figure 3.3 Boundary conditions of second gait (a) The boundary condition without weight applied on right leg single support phase. (b) The boundary condition with weight applied on right leg single support phase.

(a)



(b)

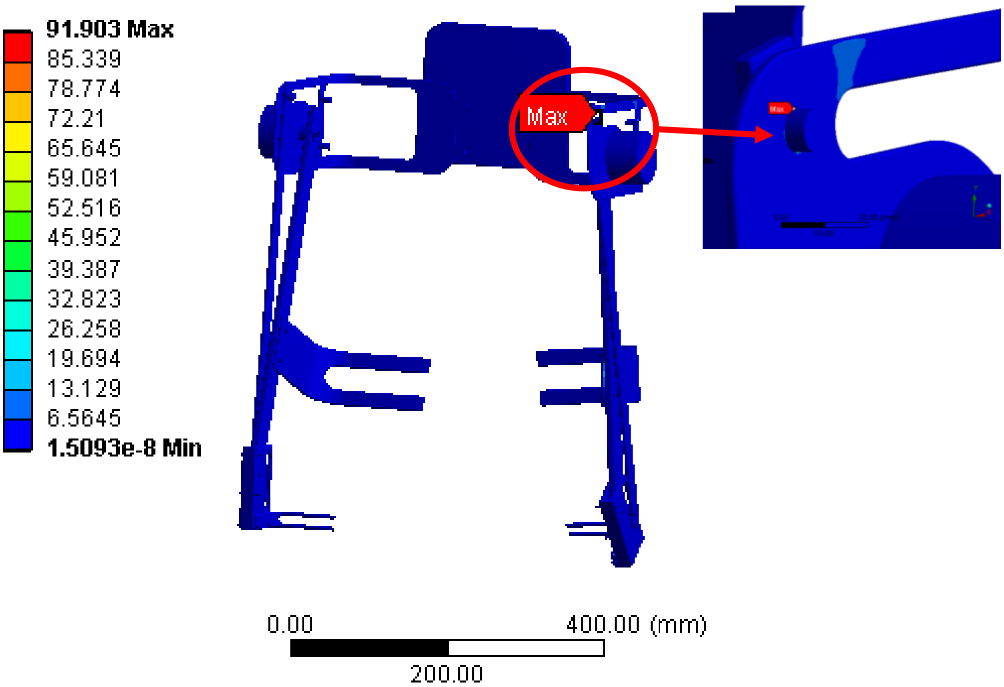


Figure 3.4 Results of second gait (a) The result and the place where the maximum stress occurs without weight applied on right leg single support phase (b) The result and the place where the maximum stress occurs with weight applied on right leg single support phase.

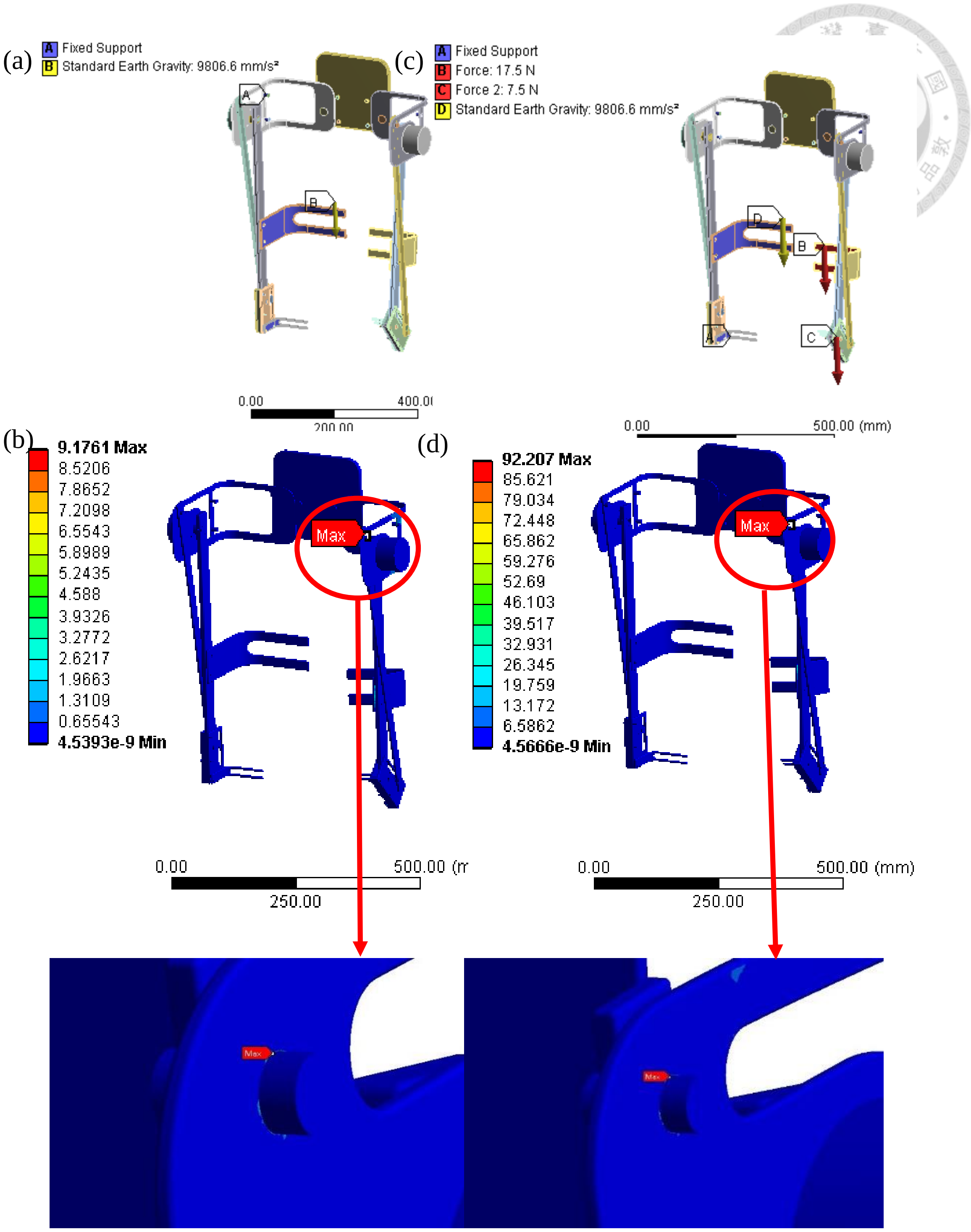


Figure 3.5 Boundary conditions and results of third gait (a) The boundary condition without weight applied on right leg single support phase. (b) The result and the place where the maximum stress occurs without weight applied on right leg single support phase. (c) The boundary condition with weight applied on right leg single support phase. (d) The result and the place where the maximum stress occurs with weight applied on right leg single support phase.

Chapter 4 Experiment process

4.1 Experiment instrument

In this section, we would introduce the experiment instrument which would be used in experiments for our assist device. Before fabricating the actually size in Chapter 2, we fabricated the simple mechanism to try to suitable size and method for fabricate conveniently. About the size we discuss in Chapter 2, but many sizes were decided by the experiment here. For example most thickness of the devices were depended on the experiment here. We tried to fabricate the suitable size but use easier method so we used wood as simple material to experimenting here. First we fabricated a bigger size of the assist device and found whether it had any problems on manufacturing or mechanism. And then we fabricated the smaller size of the assist device to solve the problem on the bigger size and make the mechanism more suitable. With the smaller size of the mechanism, we could try the limit size of our last design, and make it lighter. Because of the assist device of right leg and left leg was almost symmetrical, we only make one bigger size and one smaller size as right leg and left leg respectively.

Figure 4.1 shows the bigger size of the left leg, and the smaller size of the right leg. And then the front view, back view and side view of whole mechanism and the CAD picture of mechanism were compared as shown in Figure 4.1(a) to (f) respectively. The assist device is 2.37 kilogram without the motor, the bigger right one leg is about 1.15 kilogram and the smaller one left leg is about 0.83 kilogram. Figure 4.2 (a) to (c) shows the CAD picture of knee part, the bigger knee part and the small knee part. The weight of these component are 0.18 kilogram and 0.12 kilogram respectively. Figure 4.2 (d) and (e) shows the CAD picture of hip bar, the bigger hip bar and the hip bar. The weight of these component are 0.52 kilogram and 0.26 kilogram respectively. Figure 4.2 (f) to (i) shows the CAD of waist part and the wood device. We didn’t make right and left waist part different. Because we thought the size was suitable. The weight of this component is 0.345 kilogram. After the experiment on the wood type of the assist device, we could get the probable for the actual size for our assist device.

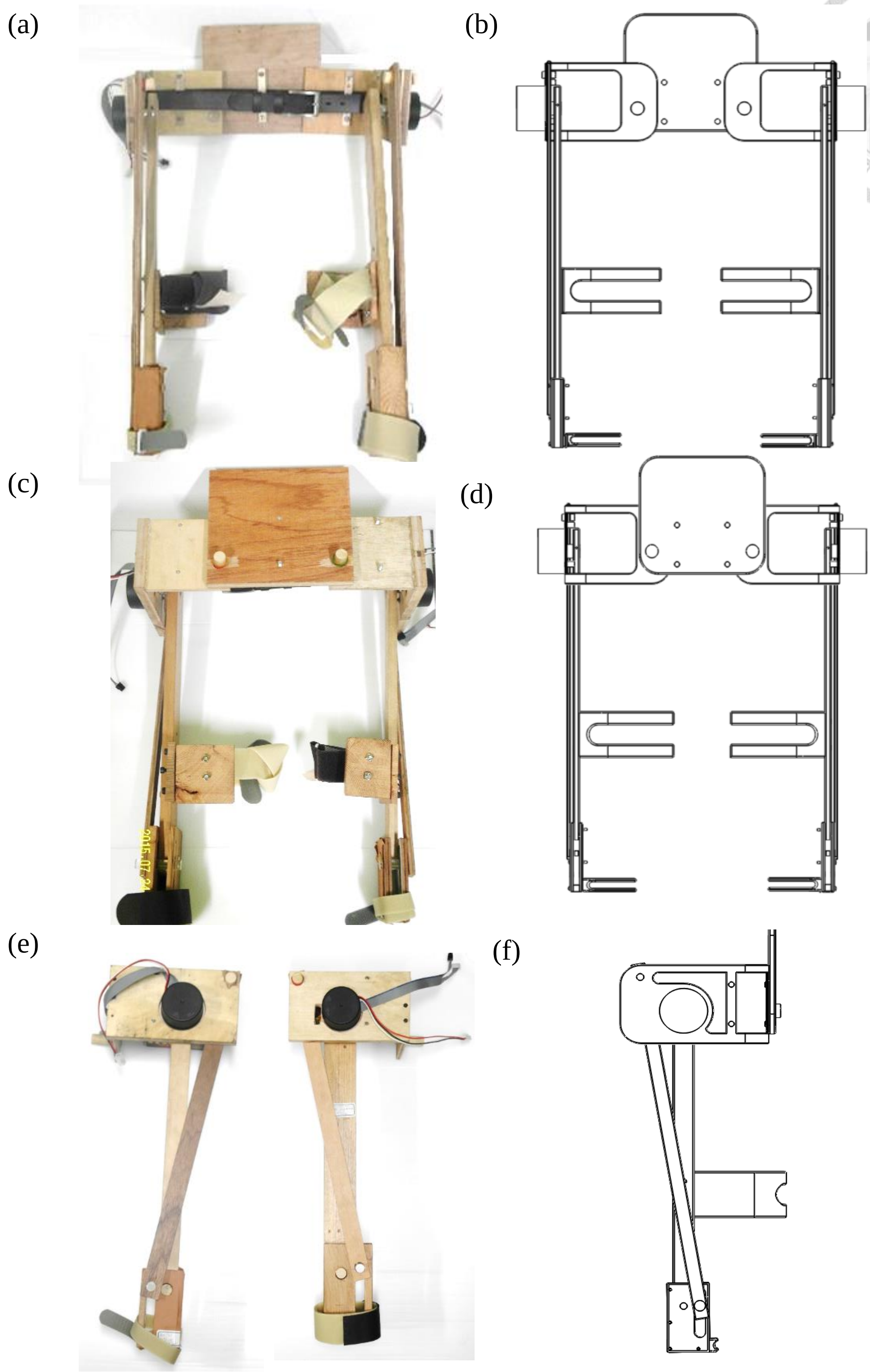


Figure 4.1 Wood type assist device and CAD picture (a) front view of wood type (b) front view of CAD picture (c) back view of wood type (d) back view of CAD picture (e) side view of wood type (left bigger / right-smaller) (f) front view of CAD picture

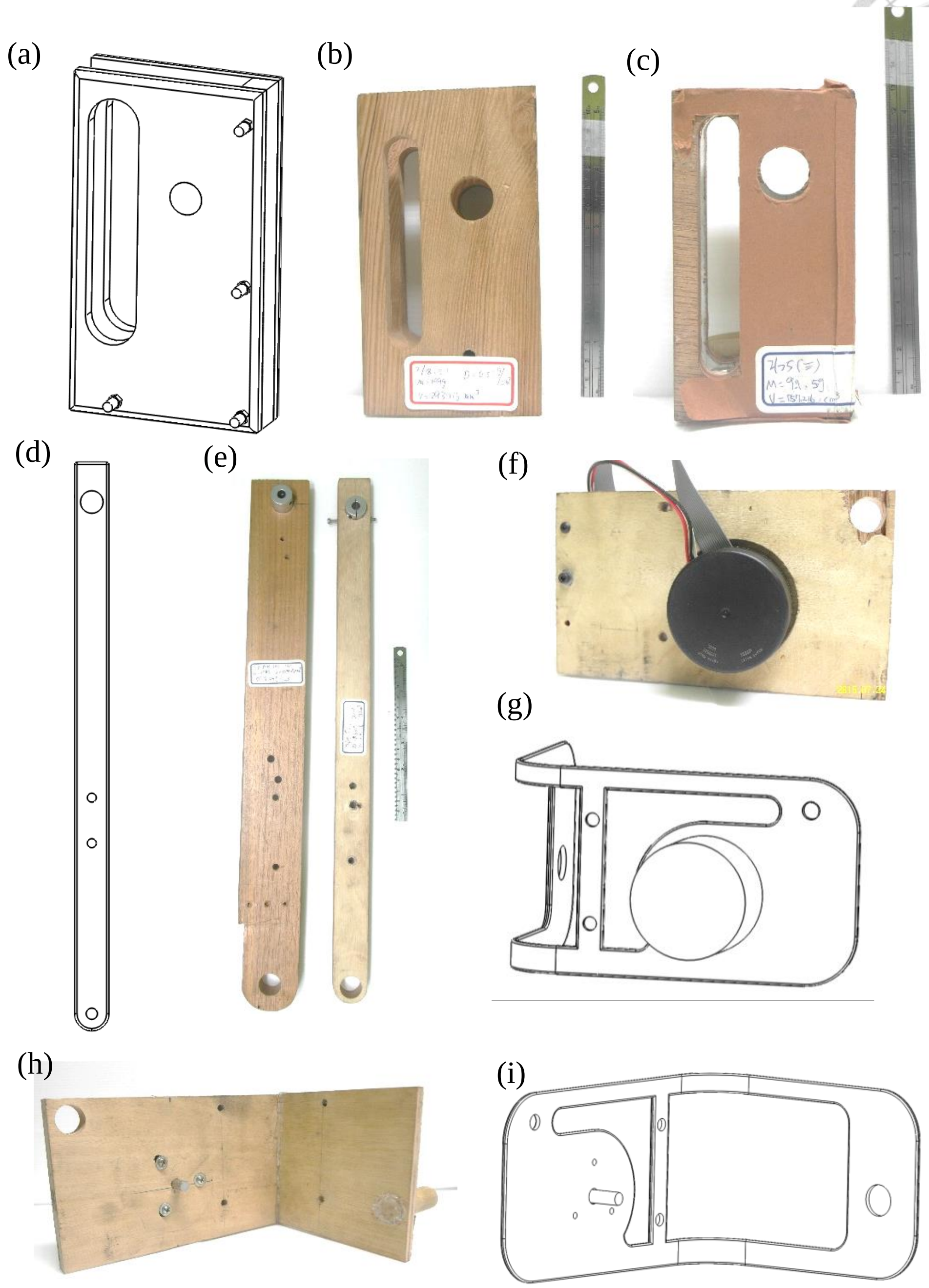


Figure 4.2 Wood type of each component and CAD picture (a) CAD picture of knee part (b) Bigger size of knee part (c) Smaller size of knee part (d) CAD picture of thigh bar (e) Bigger size and smaller size of thigh bar (f) Waist part of wood type (g) CAD picture of waist part (h) Another view of waist part (i) Another view of waist part CAD picture

Since the wood type above, we tried to fabricate the metal type by Aluminum. Figure 4.3 (a) to (e) shows the metal type of the device with the front view, back view and side view of whole mechanism. The weight of whole mechanism is about 3.2 kilogram with motor. Figure 4.4 (a) and (b) show the metal type of the knee part. The weight of this component is 0.186 kilogram. Figure 4.4 (d) shows the metal type of the thigh part whose weight is 0.388 kilogram. Figure 4.4 (f) shows the metal type of the waist part whose weight is 0.239 kilogram. The device above were fabricated by CNC and laser machining. The black color was fabricated by anodic bonding technology to make more beautiful.

Except our mechanism we also have some components on our assist device. The motor MAXON EC 60 flat which we use to control the assist device. We don't introduce the motor in detail here because we do not control the assist device in this thesis. About the axes on the mechanism, we do not choice bearing combining with cylinder to be our axes because the rotation on our assist device is not really quickly. Also we want to choice simple method to fabricate our axes. We use rustless steel shaft and C-type buckle to make our axes on the rotation of abduction and adduction on hip joint and the rotation of knee. We cut the notch to mount the C-type buckle on the bilateral of the rustless steel shaft.

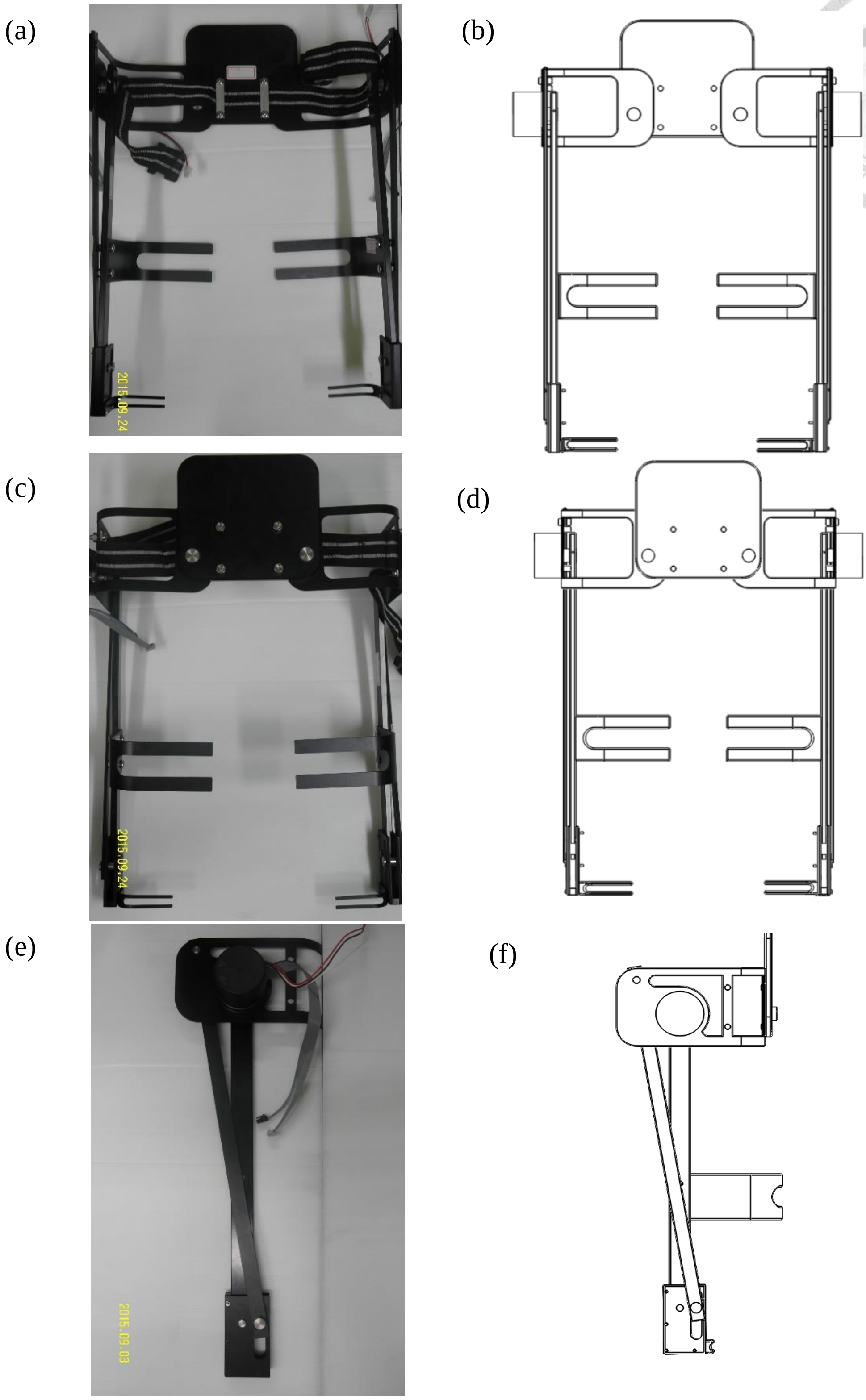


Figure 4.3 Metal type assist device and CAD picture (a) front view of wood type (b) front view of CAD picture (c) back view of wood type (d) back view of CAD picture (e) side view of wood type (left bigger / right-smaller) (f) front view of CAD picture

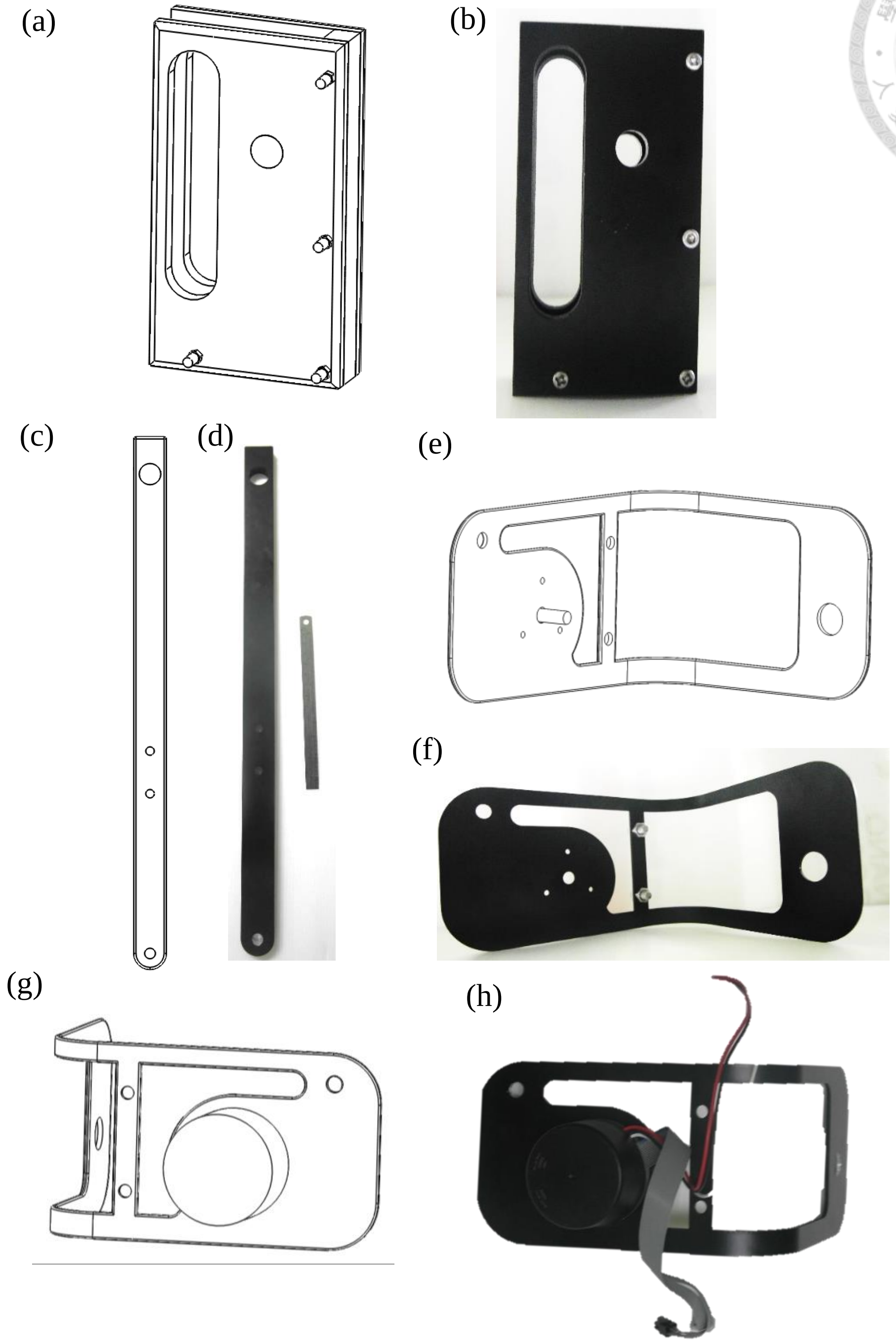


Figure 4.4 Metal type of each component and CAD picture (a) CAD picture of knee part (b) Knee part of metal type (c) CAD picture of thigh bar (d) Thigh bar of metal type (e) CAD picture of waist part (f) Waist part of metal type (g) Another view of waist part (h) Another view of waist part CAD picture

4.2 Experiment method

In this section, we want to introduce our experiment. We want to prove feasibility on our assist device. First, we use CAD to test assist device. Make the device to fit with the certain degree in normal human gait. We want to find whether it has some problems on our assist device or not. Then we fabricated the wood type for one leg and we make another smaller one to test suitable size for fabricating next metal type. Also, we test the assist device to fit with the certain degrees in normal human gait for making sure the feasibility on our assist device. Reducing most problem on our assist device to make the metal type better. And then we tried to wear the assist device, making the tester to walk with his normal gait. Measuring the degree on hip and knee joint and comparing with the degree on normal gait. To find the problem with the tester wear on. We did the experiment for right leg and left leg because the size of the right leg and left leg were not same on the wood type. Certainly, we do same experiment on metal type.

Chapter 5 Result and discussion

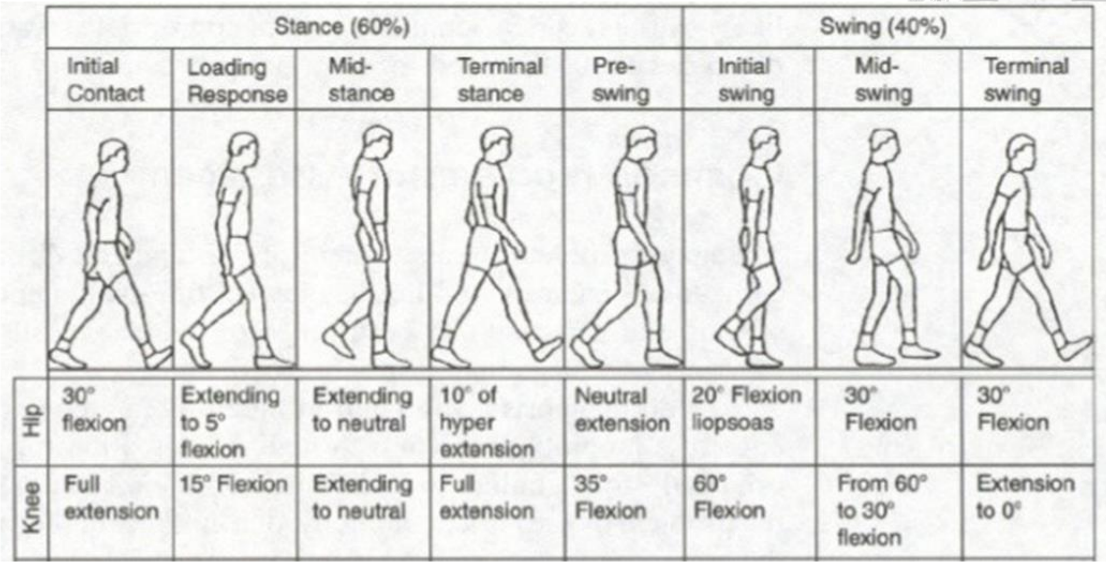
5.1 Moment of the assist device

In this section we want to prove the moment of our assist device. Whether the assist device can work or not. Comparing with the normal gait cycle in Chapter 2 and then the tester wear the assist device to find out the problem. Because we fabricated the wood type and solve the problem then fabricated the metal type. We would discuss the wood type first.

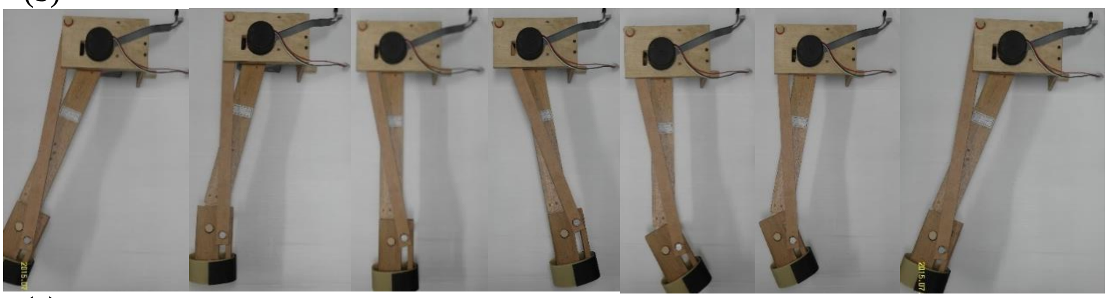
Figure 5.1 (a) shows the whole gait cycle with certain degrees. The simulation of our assist device by CAD as shown in Chapter 3 where we knew that our mechanism can fit with the gait cycle with certain degrees. So we fabricated the bigger type to make sure the suitable size. With the bigger wood type, we also make a simple experiment. Figure 5.1 (b) shows the bigger wood type with whole gait cycle and certain degrees. And then we fabricated the smaller wood type as shown in Figure 5.1 (c), because this bigger size to fabricate our metal assist device will be very heavy so we try another smaller wood type on another leg. Then we find the tester to wear the assist device and taking pictures on every gait in normal gait cycle. The tester has no previous history of musculoskeletal problems; his weight and height are 60 kg and 175 cm, respectively. Figure 5.2 shows the experiment of bigger size on left leg and smaller size on right leg with the tester wearing on.

Figure 5.1 (d) shows the metal type of the assistive device without tester wear on and correspond with the normal gait cycle. We only show the right side of our device here because the size of left side is same as right side. The tester has no previous history of musculoskeletal problems; his weight and height are 80 kg and 170 cm, respectively. Figure 5.3 shows the experiment of metal type with the tester wearing on.

(a)



(b)



(c)

(d)

Figure 5.1 Result of wood and metal type without tester (a) Gait cycle with actually degrees. (b) Moment of left leg on wood type. (c) Moment of right leg on wood type. (d) Moment of right leg on metal type.

(a)

(b)

(c)

Figure 5.2 Result of wood and metal type with tester (a) A gait cycle with the tester wear the wood assist device on left leg. (b) A gait cycle with the tester wear the wood assist device on right leg. (c) A gait cycle with the tester wear the metal assist device on right leg.

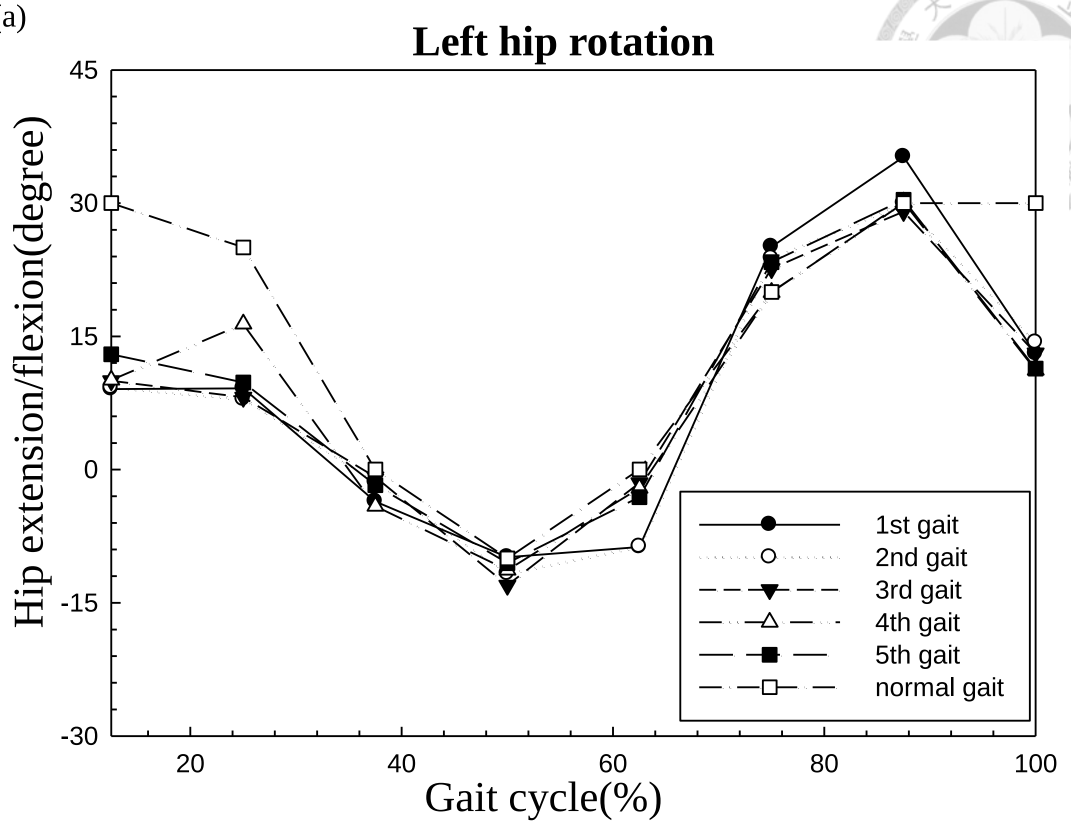
5.2 Gait cycle of the assist device

Measuring the degrees of hip joint and knee joint by ImageJ software, to find out the problem on our assist device. We did five experiments then we make the picture of gait cycle and degrees. Compare five experiments with normal human gait cycle in chapter 2 we introduced as shown in Figure 5.3. Figure 5.3(a) shows the degrees of hip joint and gait cycle of bigger size on left leg and Figure 5.3(b) shows the degrees of knee joint and gait cycle of bigger size on left leg. And one standard deviation and average value compare with normal gait as shown in Figure 5.4. Also, the result of smaller wood type can be seen in Figure 5.5 and 5.6.

The trend of our experiment were similar to the normal human gait cycle. Although the degrees have some different to normal gait cycle, some moment have large difference to average value. The problem can be traced with the wear on the thigh and calf. Although the wear on thigh can probably work, the obvious difference can be seen on the knee joint. It was because of our assist device only fabricated to knee part. During walking, the knee part can’t whole follow the rotation of calf. But the smaller size on the right leg improve the difference. According to the result, the smaller size on the right can fit with the normal human size better. So the metal type of the assist device can be fabricated in smaller size.

The results of five experiments on the metal type are shown in Figure 5.7. Normal gait is also plotted in Figure 5.7 for comparison. And one standard deviation and average value compare with normal gait as shown in Figure 5.8. The trend of the hip and knee rotation is similar to our experiment. However, the exoskeleton has relatively smaller rotation due to friction. In some situation in the gait cycle, the knee part cannot actually rotate with the calf. Although the mechanism can correspond to the normal gait, the tester in the exoskeleton cannot perform normal gait. For example, the degree at 60% to 70% of gait cycle in Figure 5.7 (b). The rotation of knee was the relative rotation between the thigh and calf. It was because of the wear on the knee, we fabricated the knee part as small as the restriction we set. But the wear on the knee can't whole drive the knee part. In some situation of the gait cycle, the knee part can't actually rotate with the calf. Although the mechanism can correspond the normal gait, the mechanism with the tester wear on can't correspond the normal gait. Finally, we ask the tester's feel when wearing the assist device. The tester said that the wear on the waist was not very comfortable and the whole assist device is too heavy for wearing.

(a)



(b)

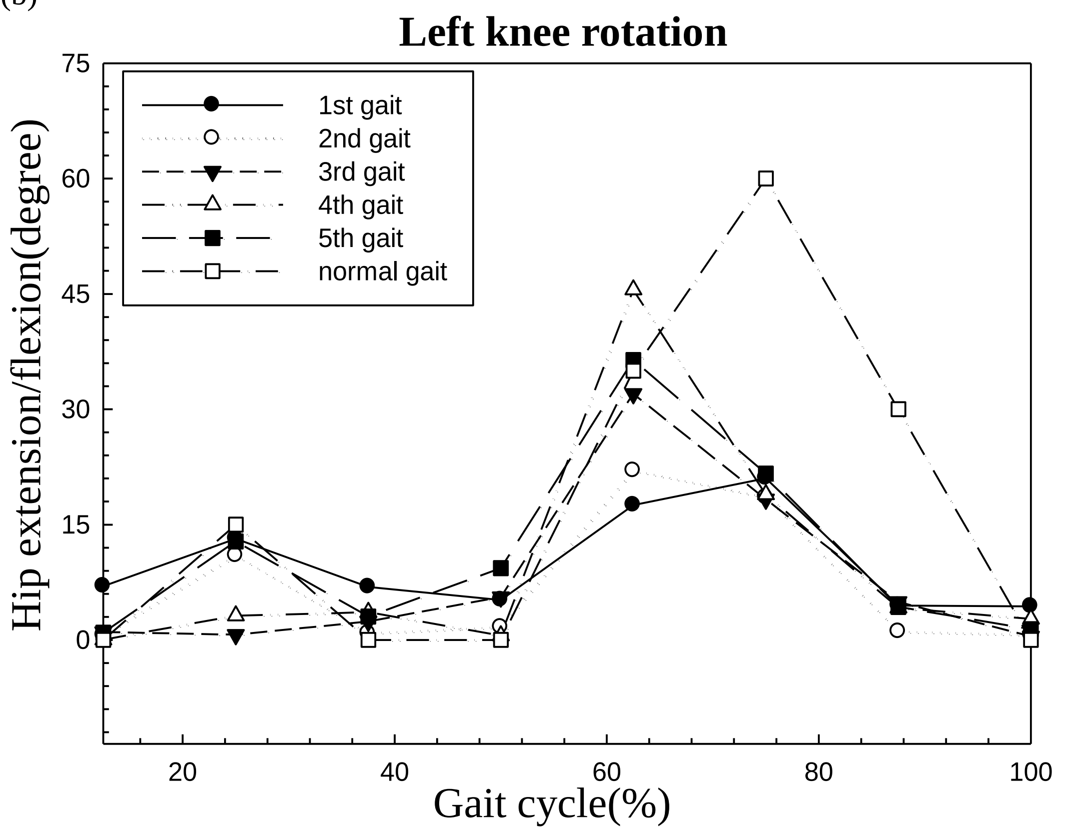
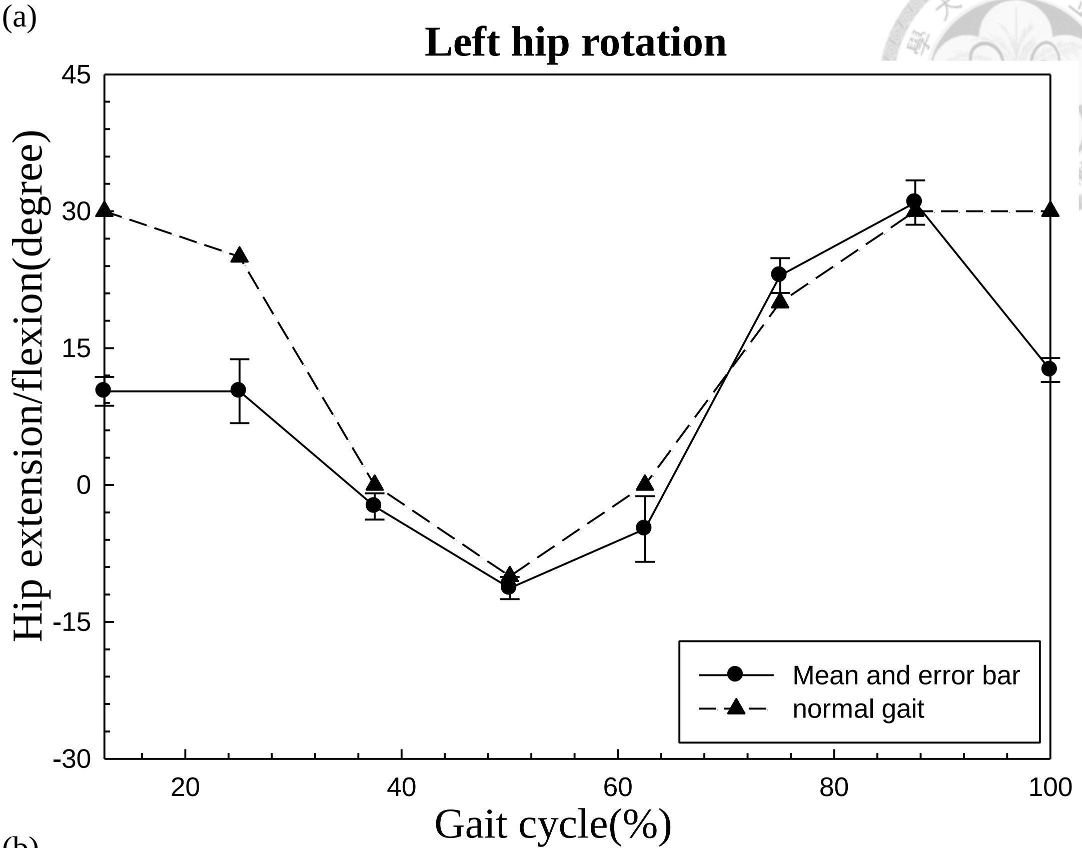


Figure 5.3 Gait cycle of the tester wear assist device (bigger wood type (left)) (a) hip rotation (b) knee rotation

(a)



(b)

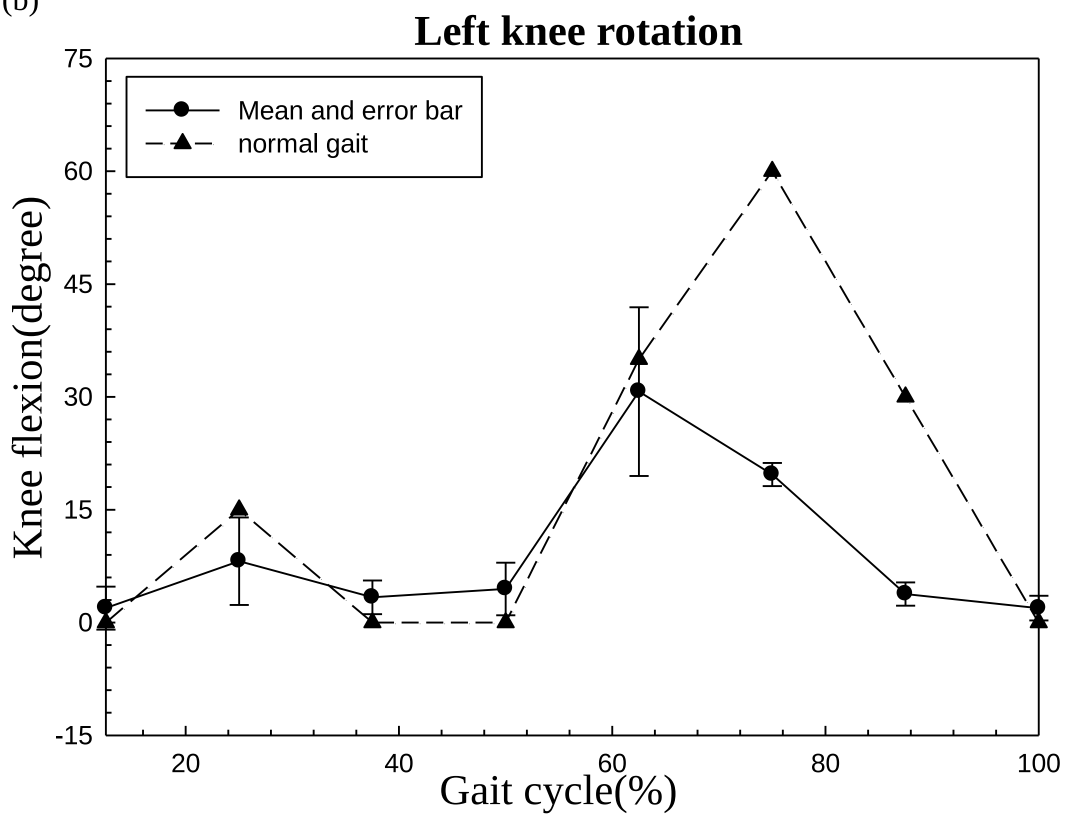


Figure 5.4 The mean and error bar compare with normal gait (bigger wood type (left)) (a) hip rotation (b) knee rotation

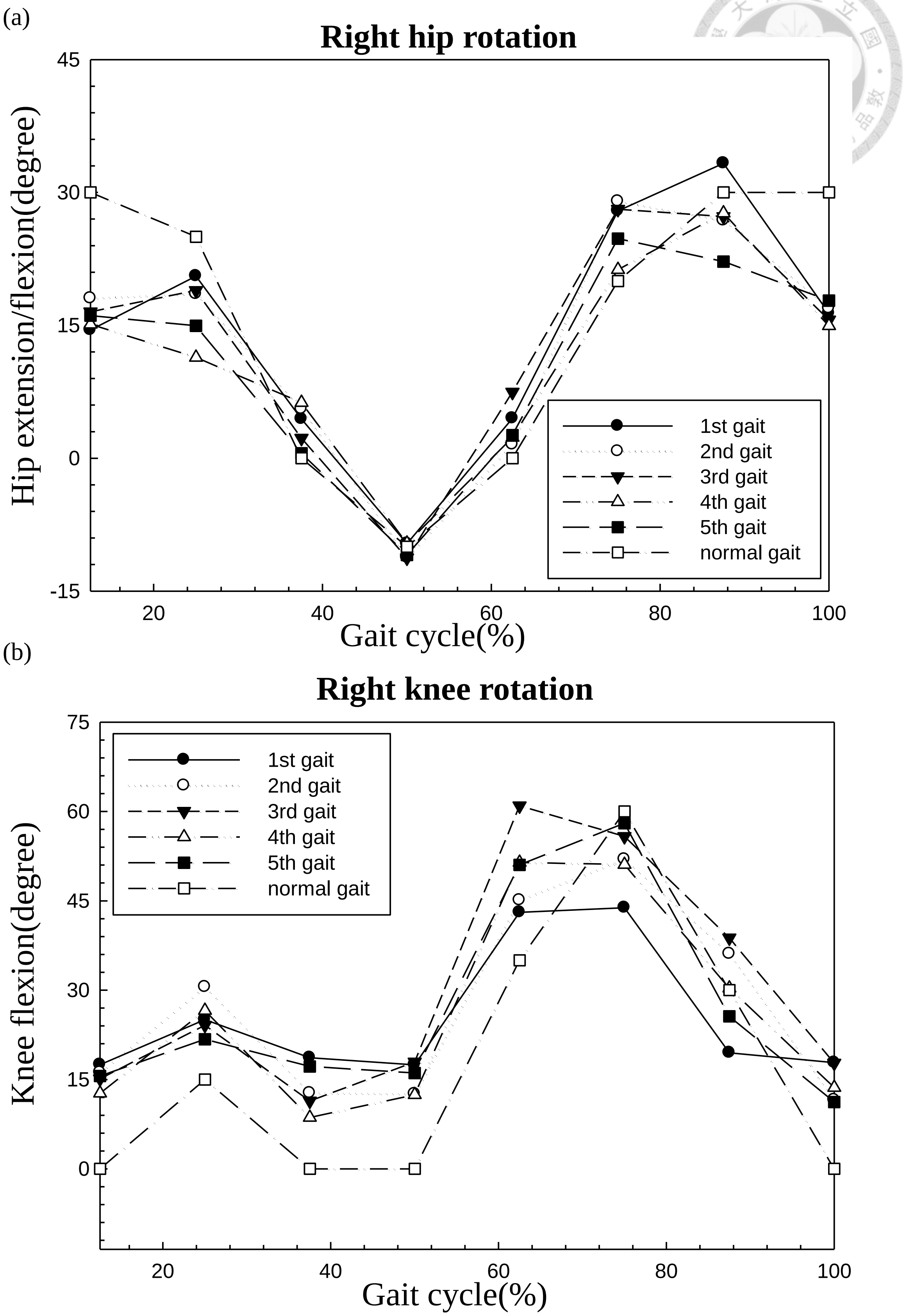
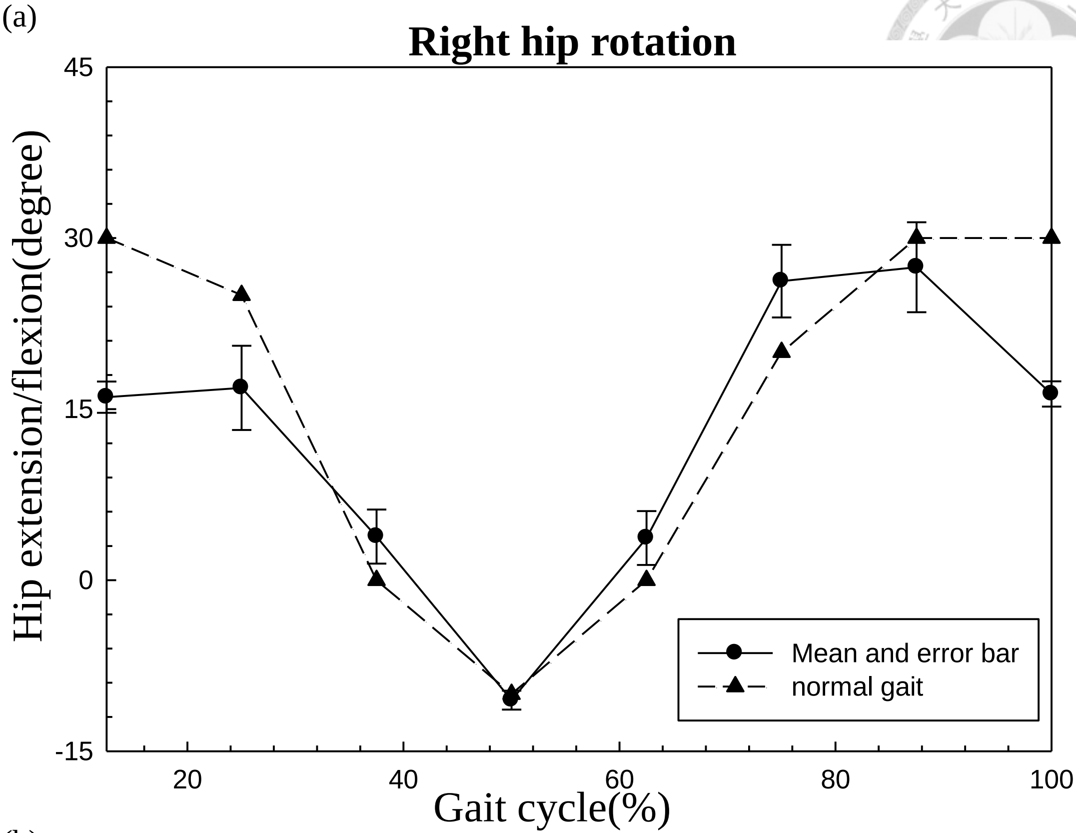


Figure 5.5 Gait cycle of the tester wear assist device (smaller wood type (right)) (a) hip rotation (b) knee rotation

(a)



(b)

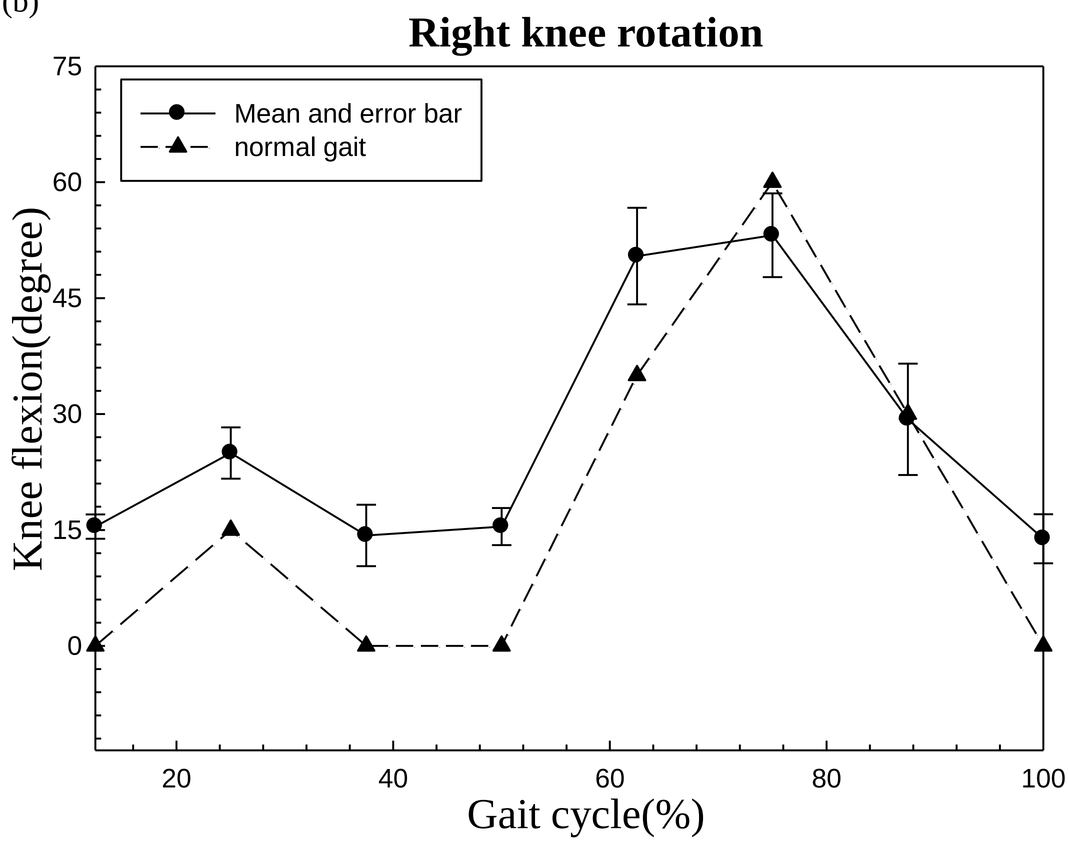


Figure 5.6 The mean and error bar compare with normal gait (smaller wood type (right)) (a) hip rotation (b) knee rotation

(a)

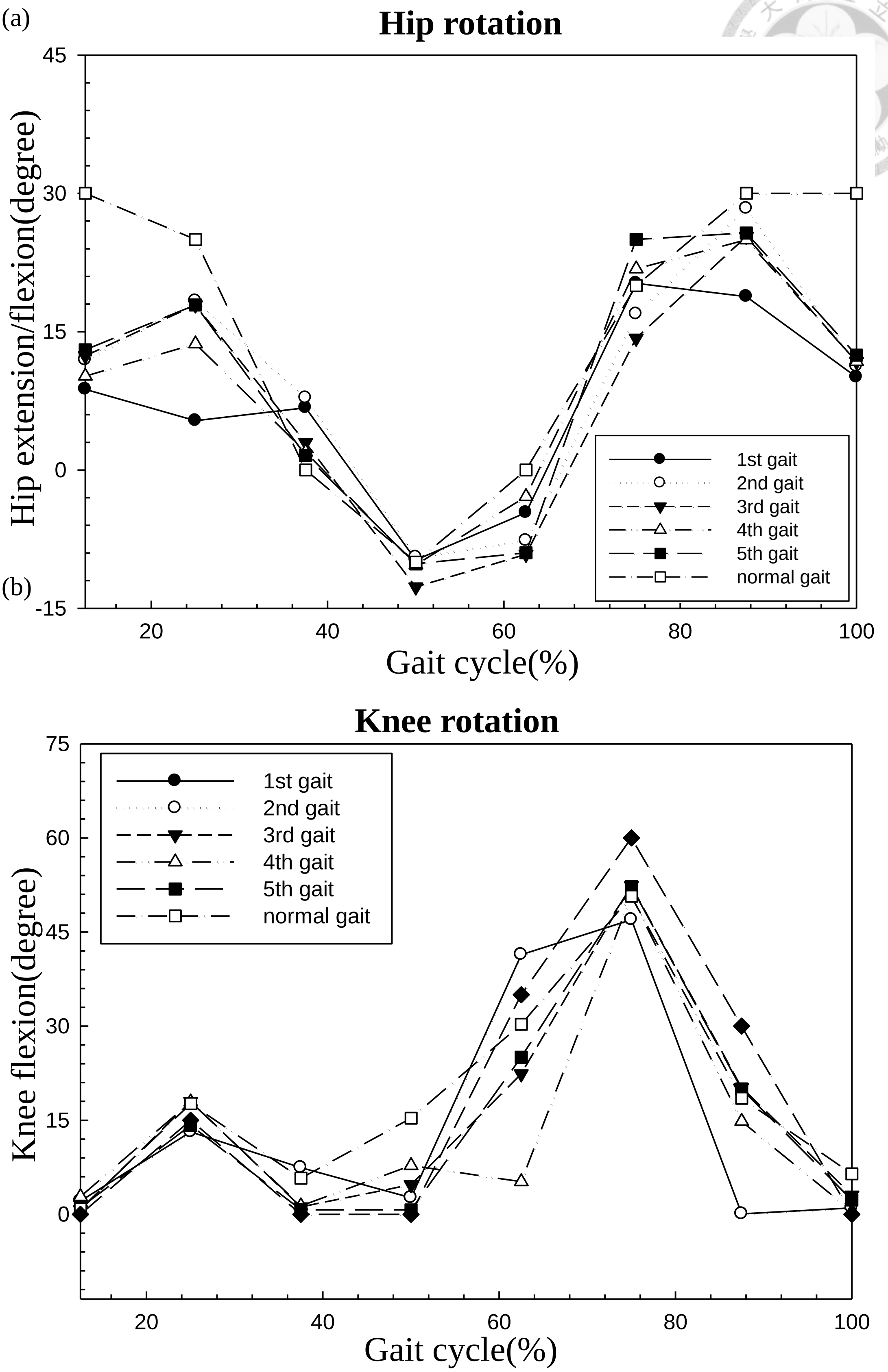
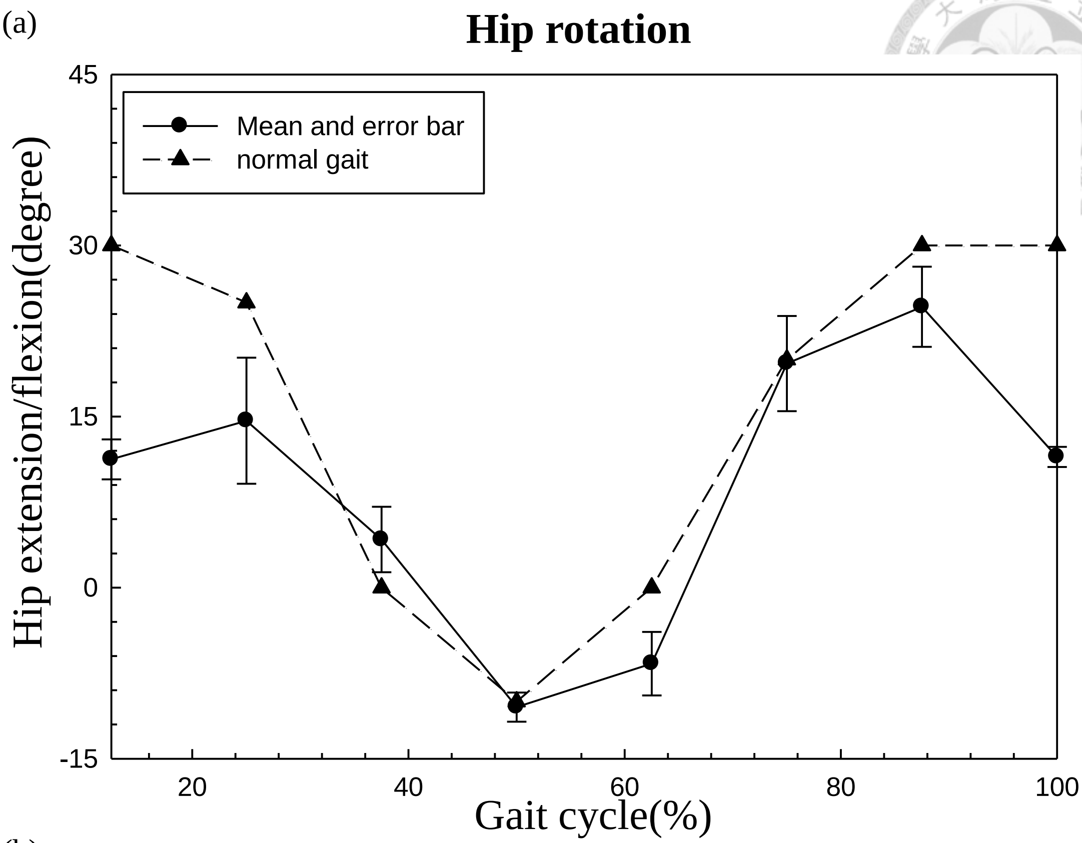


Figure 5.7 Gait cycle of the tester wear assist device (metal type) (a) hip rotation (b) knee rotation

(a)



(b)

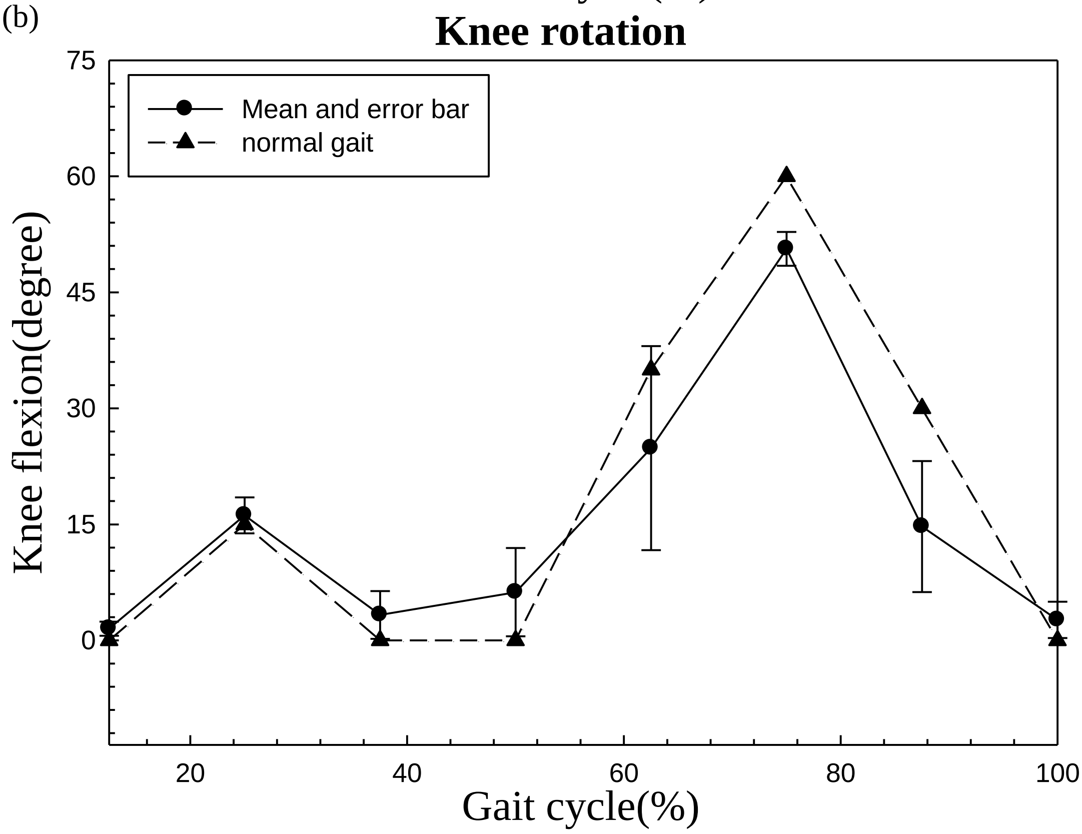


Figure 5.8 The mean and error bar compare with normal gait (metal type) (a) hip rotation (b) knee rotation

Chapter 6 Conclusions and future works

Our mechanism of the assist device can coincidence the human gait without jamming. Although it was a problem on wearing the assist device, it generally arrival our purpose. Our purpose is to design a walking assist device which can help the elderly who have the ability to walk, but can't walk like a normal person or they need a crutch to help walking. With our assist device, they are able to have their own walking assist devices so as to take care of themselves. We need to make an assist device to suit the elder and we need to make a cheaper assist device. Most assist device provided with multifunction have lots of sensors and excellent control systems with very high price. As a result, the elder need lots of money to get the device. Since there will be a population explosion in the elderly. If the cost of the assist device is very high, there must be many elders who can't afford assist devices. We use simplified mechanism and lower the cost in order to make it come true. We focused on the most important problem in here was the preventing falling mechanism. We could know the preventing falling mechanism can work as our expectation. It can restrict the rotation between thigh and calf but we can't actually know how the effect on preventing falling get. We do not quantize the effect on prevent falling mechanism. In future, there are two things we need to solve. First one is improving the wear on human's calf and waist. To make the wearer being more comfortable and feeling lightweight. Second one is need to know how is the effect on prevent falling mechanism. With these problems being solved, we can make the assist device being better and then do some control in the future. To arrival the purpose on taking care of the elderly and to solve the problem on population aging.

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