BIOMECHANICAL STUDY OF GAIT

COORDINATION OF TRANSFEMORAL

AMPUTEES

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CERTIFICATE OF ORIGINALITY

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(XU Zhi)

ABSTRACT

Transfemoral amputees' rehabilitation presents difficulties owing to the extensive loss of joints and muscles. Wearing a transfemoral prosthesis is a necessity for amputees to restore bipedal locomotion ability. Nevertheless, the prostheses still fall short of compensating the lost biomechanical function because of the scarce knowledge of amputees' gait coordination.

Therefore, the overall objective of this study was to better understand the gait coordination of transfemoral amputees and its implication in rehabilitation and prosthesis design. The scope of the study included investigating transfemoral amputees' gait coordination of lower limbs, and studying the effects of the prosthetic knee–ankle coupling designed mechanism and walking speed on gait coordination patterns.

Our first study investigated the gait coordination of unilateral transfemoral amputees and compared to abled persons. The results suggested that amputees adopt a different gait coordination pattern to achieve the specific prosthetic gait requirements. The gait coordination analysis demonstrated its sensitiveness in detecting gait deviation by studying the coordination between joints rather than studying each joint separately.

The second study compared the gait coordination of transfemoral amputees wearing traditional prostheses and the knee–ankle coupling designed prosthesis. The latter prosthesis guaranteed better gait compliance through ankle dorsiflexion and plantarflexion, while the prosthetic knee showed a function similar to that of the

traditional prostheses. The coupling designed mechanism exhibits advantages in improving gait coordination, but there is still room for improving the prosthetic knee extension function to induce a better gait.

In the third study, transfemoral amputees walked at different speeds. The gait coordination was less adjustable in response to slow speed, which gives us another explanation for the amputees' low walking speed. The preferred speed was the "optimal" speed at which the motion demonstrated higher flexibility. The prosthetic knee and ankle displayed weak adaptability to different walking speeds, whereas the hip joints played an essential role for control adjustment when coping with different speeds.

The fourth study compared the muscle coordination between non-amputated limbs and the limbs of abled persons through inverse dynamics simulation. The realignment of muscle coordination pattern was the main means to satisfy the specific joint moment requirement of non-amputated limbs. In addition, transfemoral amputees were found to hyperextend their intact knees during stance, which may be the cause of knee osteoarthritis. We also compared the maximum forces of each muscle and found muscle disuse in non-amputated limbs.

In conclusion, the gait coordination of transfemoral amputees differs from that in abled persons. These differences in gait coordination embody deficiencies in motion control ability along with defects of prosthesis. The gait coordination analysis demonstrated advantages in its multi-level research, namely from joints to joints and muscles to muscles, thus it is more sensitive in detecting gait deviation. The findings could be beneficial in interpreting the secondary pathology of transfemoral amputees and giving another insight into the ways of improving

rehabilitation programs and prosthesis design.

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LIST OF ABBREVIATIONS

GC： Gait cycle

BW: Body weight

GRF: Ground reaction force

SACH: Solid ankle cushion heel

ESR: Energy-storing-and-releasing

HS: Heel strike

IC: Initial contact (IC)

LR: Loading response (LR)

MSt: Middle stance

TSt: Terminal stance

PS: Pre-swing

IS: Initial swing

MSw: Middle swing

TSw: Terminal swing

CRP: Continuous relative phase

CC: Cross-correlation

RMS: Root mean square

DP: Deviation phase

DI: Decomposition index

BFL: Biceps femoris caput longum

ST: Semitendinosus

SM: Semimembranosus

AM: Adductor magnus

GMa: Gluteus maximus (GMa);

ADL: Adductor longus

RF: Rectus femoris

GRa: Gracilis

Sar: Sartorius

ILi: Iliacus (ILi);

VL: Vastus lateralis

VM: Vastus medialis

VI: Vastus intermedius

BFB: Biceps femoris caput breve

Pop: Popliteus

GAs: Gastrocnemius

SOl: Soleus

TP: Tibialis posterior

FDL: Flexor digitorum longus

FHL: Flexor hallucis longus

Per: Peroneus

TA: Tibialis anterior

EH: Extensor hallucis

ED: Extensor digitorum

CHAPTER I INTRODUCTION

A large number of amputees are currently living worldwide. Compared to other types of lower extremity amputations, transfemoral amputations are more difficult to rehabilitate owing to the significant loss of joints, bones, and muscles. Wearing a prosthesis has become the most important way to restore the bipedal locomotion of transfemoral amputees, and it is also part of the physical therapy. Additionally, it is beneficial to psychotherapy owing to the cosmetic appearance.

1.1 Amputation prevalence

Amputation, which refers to the removal of all or part of the limb, is an operation that has to be performed to save or prolong the lives of the wounded and sick. It aims to amputate limbs that have lost viability, endangered lives or without physiological function. The common causes of amputation are severe trauma, infection, tumor, diabetes mellitus, and its complications (Hagberg and Brånemark, 2001; Mutlu et al., 2011; Pillai et al., 2011; Sagawa et al., 2011; Wong, 2005). In addition, all kinds of peripheral vascular occlusive diseases related to blood and vascular system (such as arteriosclerosis obliterans, thromboangiitis obliterans, arterial embolism of lower extremity, *etc.*) may induce lower extremity lesions, which have ultimately to be amputated for the gangrene of lower extremities (Blevins and Schneider, 2010; Piazza and Creager, 2010). It is worth mentioning that the amputation rate in diabetic patients is 46 times higher than in non-diabetic people (Soomro et al., 2013). World widely, amputations owing to diabetes account for 85% of lower extremity amputations, followed by trauma and neoplasms (Soomro et al., 2013). In Hong Kong, China, infection, vascular diseases and

trauma accounts for 35%, 31% and 26% of amputations, respectively (Wong, 2005). It is figured out that in part of developing countries, mine-induced injuries account for as much as 80% of amputations (Berry, 2003; Collins et al., 2006; Islinger et al., 2000; Meade and Mirocha, 2000) and that more than 300,000 people have been injured or amputated owing to mines (Network, 2003). Krueger *et al.* studied amputation trends during the wars between Iraq and Afghanistan from 2001 to 2011 and counted 1,631 amputation, which accounted for 3.6% of total trauma over a decade (Krueger et al., 2012).

Of the amputations resulted from vascular diseases, about 97% are lower limb amputations, of which 25.8% are transfemoral amputations (Pillai et al., 2011). In the United States, approximately 79.5% of amputees suffered from lower extremity amputations (Pillai et al., 2011). In the wars between Iraq and Afghanistan, the main amputations were transtibial amputations (41.8%), followed by transfemoral amputations (34.5%). Approximately 30% experienced mixed amputations (Krueger et al., 2012), and most which were males (Chan et al., 1984). Amputations increased in 2010 and the first half of 2011, and the incidence of multiple amputations was higher than previous. Other studies indicate that the incidence of traumatic amputations for military reasons is on the rise in 80 countries (Network, 2003). Although the frequency and types of amputations vary from country to country, the number of amputations has increased significantly in the last century due to wars and the development of the automotive industry.

In recent years, earthquakes have befallen frequently all over the world, about one million times a year, which is equivalent to about two times a minute. More than 780,000 people died for disasters, with earthquakes accounting for about 60% of all

disaster deaths and approximately 2 million people directly affected by earthquakes (Bartels and VanRooyen, 2012; Naghii, 2005). For the past years, ever more people in China have had their limbs amputated by earthquakes (Feng and Zhang, 2009; Jin-hua, 2009; SU et al., 2009; Zhuang and Hu, 2008). According to the *World Health Organization*, the Wenchuan earthquake instigated a total of 69,197 deaths, 374,176 injuries and 18,238 disappearances. Moreover, there happened to be 46.26 million people affected, of whom 374,176 injured. 39 people amputated at the *West China Hospital, Sichuan University* after the earthquake (XIE et al., 2008); in addition, the *Third Military Medical University* admitted 149 patients injured by the earthquake, of whom 9 with limbs amputated and 7 with lower limbs amputated (Feng and Zhang, 2009).

Owing to the wound healing and muscular atrophy of the residual limb, it may induce vascular occlusion, embolism, and obstruction of blood supply, which may result secondary or multiple amputations (Shapiro and Huang, 2009). Amputees also face severe consequences such as a high incidence of complications (34.7%) and a 30-day mortality rate of 7% (Jr et al., 2011). Not only do residual limbs embrace higher pathological problems, but also amputees with non-amputated limbs and other parts of the body also present severer symptoms in the process of rehabilitation. For example, more than 71% of unilateral amputees manifested back pain and non-amputated limb pain (Burke et al., 1978; Ephraim et al., 2005; Morgenroth et al., 2010; Norvell et al., 2005; Struyf et al., 2009).
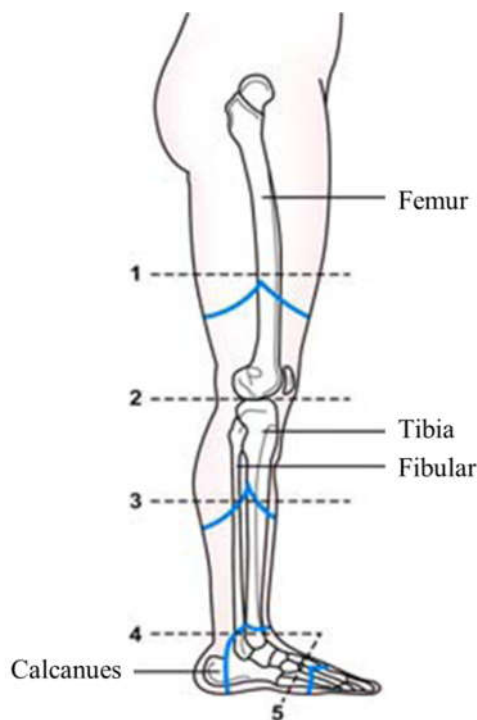
Fig. 1-1 Levels of lower limb amputation (Mutlu et al., 2011)

One of the basic principles of amputation is to preserve the length of the limb as long as possible while meeting the requirements of surgical treatment. The more limbs are amputated, the more bones, muscles and joints are lost, and the greater the impact, the heavier the burden. For some special parts of amputation, the amputation plane design also needs special treatment to reserve space for other joints and limbs. For example, amputation of the lower leg should take place at the junction of the lower and middle third of the leg. Depending on the amputation location, lower limb amputations can be of five levels (as Fig. 1-1):

1) Transfemoral amputation;

2) Knee disarticulation amputation;

3) Transtibial amputation;

4) Ankle disarticulation amputation;

5) Transmetatarsal amputation.

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Nowadays, the demand for prostheses keeps increasing. According to the investigation, 88.4% (2,021 people) of the injured patients in 21 towns of Mianzhu area need physical therapy after the Wenchuan earthquake. In the United States, the need for lower extremity amputations and prostheses has increased as a result of the war on terror, resulting in more traumatic patients (Collins et al., 2006). The growing number of amputations in developing countries has led to higher requirements for the number of prostheses as well as for manufacturing capacity. A lot of researches are also devoted to the development of low-cost, functional prosthetic products. Still, the current prostheses are not enough to meet the amputee's needs. There are a large number of amputees in China. According to the data of *The Second China National Sample Survey on Disability* in 2006, there are 2.26 million amputees among the 24.12 million people with physical disabilities. Of which 630,000 are in urgent need of prostheses (Luo and Sun, 2009). Not only the number of amputees in China is huge, the amputations in foreign countries are also very serious. Following the statistics, there were 1.6 million amputees in the United States by 2005, and the number is anticipated to increase to 3.6 million by 2050 (Ziegler-Graham et al., 2008).

Although the causes, types, and frequencies of amputations vary from country to country (Wan-Nar Wong, 2005), the number of amputations has increased significantly from the last century to the present due to wars and the development of the automotive industry (Torrealba et al., 2008).

1.2 Pathology and burden of transfemoral amputees

The rehabilitation process after amputation accompanied by many problems. These pathological problems occur not only in residual limbs, but also in non-amputated

limbs and trunks with a higher incidence than in normal persons, such as:

    a)   Lower back pain (Devan et al., 2012)

    b)   Phantom pain (Ephraim et al., 2005)

    c)   Hip and knee osteoarthritis of non-amputated limbs (Struyf et al., 2009)

    d)   Decreased bone mineral density, joint degeneration (Sherk et al., 2008)

    e)   Skin problems such as vesicles, edema and pressure ulcers in residual limbs (Lyon et al., 2000; Sanders et al., 1997)

    f)   Soft tissue injury (Portnoy et al., 2009)

    g)   Muscular atrophy (Fraisse et al., 2008; Jaegers et al., 1995; Schmalz et al., 2001)

Amputees will face complicated postoperative problems and complications with the residual limb. After wearing prostheses, the interface stress between the socket and the skin will initiate the skin microcirculation to be blocked, which will lead to skin problems such as blisters, edema, pressure sores and so on (Lyon et al., 2000; Sanders et al., 1997); high stress at the osteotomy end can activate deep soft tissue damage (Portnoy et al., 2009); also, muscular atrophy of the residual limb (Fraisse et al., 2008; Schmalz et al., 2001), and so on. These problems seriously affect the rehabilitation process and quality of life of amputees, and bring a lot of burden to amputees.

Although lower limb prostheses allow amputees to regain the ability to walk upright with both feet, they are still far from natural human limb function. A series of defects still plague the amputee's daily life (Hagberg and Brånemark, 2001). Amputees need to increase the net joint moment and joint power of the healthy limb in gait compared to normal subjects, a mechanism known as compensatory (Nolan

and Lees, 2000). The additional stress on the healthy limb is detrimental to the amputee and can cause pain and even joint degeneration (Burke et al., 1978); normal stresses and shear forces acting on the interface between the socket and the residual limb can also cause pain, skin problems and gait deviation in the residual limb (Pitkin, 1997).

1.3 Prostheses and rehabilitation

After amputation, the installation of prostheses has become the main means of compensating the biomechanical function and restoring the ability bipedal locomotion. At the same time, wearing prostheses can help to restore appearance and play an important role in amputees' psychology, which is the main rehabilitation method for amputees. With the improvement of industrial technology and the progress of scientific research, prosthetic products have developed rapidly in recent decades, and more and more amputees begin to wear prostheses for rehabilitation. However, the improving standard of living also encourages amputees more and more demanding for prostheses. Therefore, Prosthetics has gradually become an important part of Rehabilitation Engineering. Amputees' requirements for prostheses are not only reflected in function and comfort, but also consume higher requirements for portability and durability, appearance and so on. Thus, Prosthetics has made a series of achievements in basic theory, structural design, material application, manufacturing and assembly technology.

The daily lives of amputees are severely affected by gait (Van Velzen et al., 2006), professional status   (Burger and Marincek, 2007), social relations (Williams et al., 2004), participation (Gallagher et al., 2011) and emotion (Singh et al., 2009). Additionally, the effects of adaptation to amputation are not only physiological but

also physical (Coffey et al., 2013).

China has always attached great importance to the production and development of prostheses. The *National High-tech Research and Development Program* (863 Program) and the *National Science and Technology Support Program* have set up prosthetic-related subjects, and the *National Natural Science Foundation of China* also encompassed prostheses in the *Strategic Report on the Development of Mechanical Engineering* (2010-2020).

Prosthesis is the interface of the human body and ground, so it undertakes an important biomechanical function. Depending on the amputation position, motor ability, and amputee' requirements, various prosthetic products are needed for amputees. For example, if amputees want to carry with a beautiful appearance, they can choose the prosthetic accessory products, such as the cosmetic shell. But for the majority of amputees, they are more eager to combine various factors to choose the traditional prosthetic parts, and wear them on the residual limbs in combination with the socket. Due to the progress of industrial technology and the development of material science, the existing transfemoral prostheses have been more humanized and comfortable than the previous products.

1.3.1 Transfemoral prosthesis structure

Typical transfemoral prostheses include the following main components (as shown in Fig. 1-2):

1) Socket

2) Prosthetic knee

3) Prosthetic ankle (including foot)
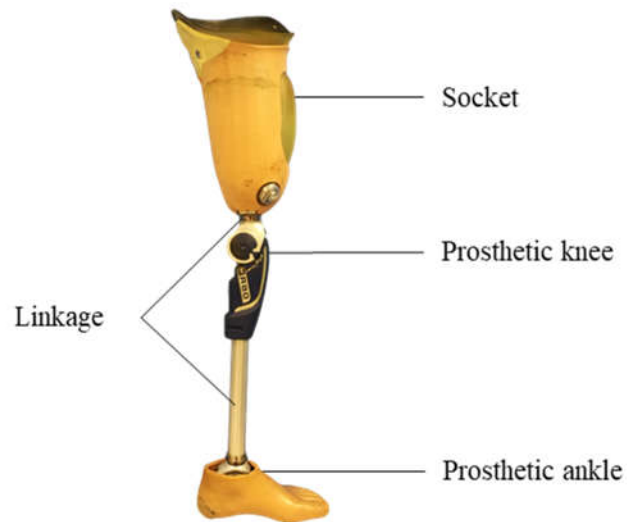
Fig. 1-2 Typical construction of transfemoral prosthesis

The socket is responsible for connecting and transferring loads, which is in direct contact with the residual limbs and forms a human-machine interface between residual limbs and prosthesis. The quality of a socket directly affects the wearing experience of amputees, and then affects the function and use of prostheses. During walking, the socket will bear the maximum load of about 120% of the body weight; at the same time in the swing phase need to ensure the stability of the prosthesis hanging on the residual limbs, which inhibits excessive relative displacement. Too tight connection between the socket and the residual limb will lead to extra pressure and skin problems of the residual limb; while too loose connection will lead to unstable suspension and large relative slippage. The loading capacity of residual limb soft tissue is relatively low, and the interface stress and distribution pattern of the human-machine interface of the socket have a direct impact on the comfort and safety of wearing. Improperly designed socket can lead to irrational stress distribution, which can initiate skin or soft tissue pain or even injury of the residual limb. Therefore, the design of the socket must take into account the various carrying capacities of the residual limb and be tailored to the specific conditions of each

amputee's residual limb (Lee et al., 2005; Zhang and Lee, 2006).

There are more and more lower limb prostheses in the market. From the mechanical single-axis and multi-axis knee joints, which can only play the most basic supporting and walking role, to the pneumatic, hydraulic, and even microprocessor-regulated intelligent prosthetic knee, ankle, prosthetic products are constantly improving in symmetry, balance and reducing metabolic energy consumption. The number of tasks that can be accomplished functionally is also increasing. These products continue to solve the defects of the past prosthetic products, such as wearing more comfortable, labor-saving, more complete functions, to bring amputees a better experience.

1.3.2 Prosthetic knee joint

The prosthetic knee joint embraces two main functions: 1. simulate the normal knee joint in the swing phase of stable extension or flexion; 2. ensure stability in that stance phase. The prosthetic knee is more complicated than the other parts because it is the connecting part between the thigh and the shank.

The prosthetic knee can be divided into uniaxial knee and multiaxial knee following the axial shape of the joint. Uniaxial knee joint consumes only a single axis of rotation. This kind of prosthesis has good flexibility in the swing phase, but the stability of the stance phase is not enough. The use of groups was biased towards sports exercise, such as young adult amputees who have higher physical activity and had better residual limb control (Liu and Diao, 2006). The multi-axis knee joint is relatively more complex, mostly using multi-link mechanisms, such as the six-bar linkage, whose structure is closer to the physiological characteristics of knee

joint movement of abled persons. The center of rotation of the six-bar linkage adjusts its position with different phases in the gait so that the stance phase remains stable and the swing phase is flexible. It can be designed flexibly according to different biomechanical characteristics.

According to the damping control method, the prosthetic knee joint can be divided into mechanical, pneumatic, hydraulic, and intelligent knee joint, as shown in Fig. 1-3.

(a)   (b)   (c)   (d)

Fig. 1-3 Classification of prosthetic knee joint by damping control： (a) mechanical knee joints; (b) pneumatic knee joint; (c) hydraulic knee joint; (d) Intelligent knee joint (Pillai et al., 2011)

Mechanical knee joint usually adopts knee lock or load-bearing self-locking device to control the stability in stance phase. The prosthetic knee does not flex after locking, thus ensuring that the lower extremities move the center of the body forward stably during the gait stance phase or standing. Load-bearing self-locking devices generally rely on friction to restrict joint movement. Unlike a knee lock, it remains stable at a certain flexion angle. However, if the self-locking devices do not bear enough weight, the stability will be affected. The control of mechanical knee joint in the swing phase mainly depends on sliding friction damping

adjustment. The friction force of the knee joint will not change with the speed. Therefore, it is necessary to manually adjust the damping magnitude for different walking speeds.

Pneumatic knee joint, which differs from mechanical knee joint, lies in the difference of dampers. The pneumatic knee joint has a pneumatic cylinder connected to both ends of the knee joint, and the piston moves back and forward during walking. The stronger the gas compressibility, the faster the piston travels after wearing; or the tighter the air opening of the flow control valve is adjusted, the greater the cushioning damping of the piston. Because of the compressibility of the gas, it is difficult to ensure the stability of the stance phase only by the damping of the pneumatic cylinder, so it is usually designed with a multi-link mechanism.

The damper of the hydraulic knee joint is similar to that of the pneumatic knee joint. The difference is that the liquid can hardly be compressed, so the liquid damper does not have preloading function like a cushioning medium of the pneumatic damper. On the other hand, the incompressible liquid also makes the damping adjustment ability and the control ability during supporting more stable and reliable. Compared with pneumatic knee joint, the technology of hydraulic damper is more complicated and difficult to adjust. These prostheses are more likely to be applied in amputees who are more athletic.

Intelligent knee is a prosthetic knee with intelligent adjustment function, as its name implies. The above-mentioned joint damping cannot be adjusted actively at any time during walking, so the adaptability is poor for different walking speeds. When the speed changes, the metabolic consumption of the amputee increases, and the amputee is prone to fatigue. Intelligent knee joint uses sensors to detect the gait

biomechanical parameters during walking. Through microprocessor analysis and then adjust the control joint damping, intelligent knee gives different gait characteristics to different control, which is more adaptable. So that the gait of the amputee is more natural and easier and labor-saving. In the realization of damping control, pneumatic cylinder or hydraulic cylinder is often used to intelligent knee joint. There are also products that use magnetorheological control. At present, passive knee joint is the main type of intelligent knee joint applicated in clinical. Amputees' gait is usually analyzed by monitoring joint angle, joint moment, and joint acceleration.
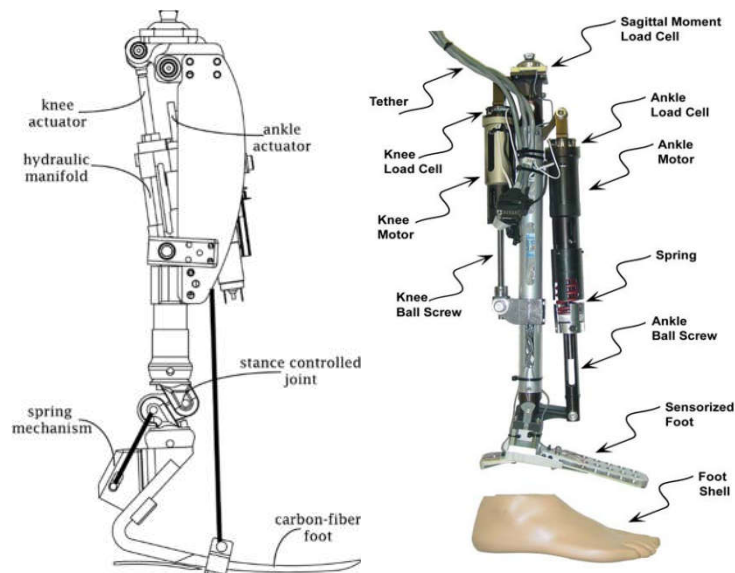
Fig. 1-4 Two active transfemoral prostheses designs (Pillai et al., 2011; Sup et al., 2008)

The active prosthetic knee is a higher-level version of the above-mentioned intelligent knee, which typically consumes electric energy to actively generate moment for the knee, driving the knee to flex or extend (Heins et al., 2018; Rohani et al., 2017). In addition to the advantages of passive intelligent knee joint, the prostheses provide energy actively to make up for the function of active extension

and flexion of the joint lost by amputation. Therefore, the prostheses can provide auxiliary force in complicated road conditions such as climbing up the slope and climbing buildings, making it easier for amputees to walk. But these products are not yet mature enough. Power supply is one of the bottlenecks that cannot be broken through. Electricity and battery, the volume and weight of the whole mechanism are all urgent problems to be solved. In recent years, many researches design the prosthetic knee in a manner that compromise the active and passive, so that the driving mode and the non-driving mode work together. This new mode not only provides auxiliary force when a large moment is needed, but also saves more energy than the pure active control structure, so it has become a new research hotspot of intelligent prosthetic knee joint. However, whether active or hybrid, the product structure is more complex. The usage and maintenance are more cumbersome than other passive prostheses, the cost is much higher too. Therefore, these products are still in the experimental stage and have not yet been widely applied. A typical design is shown in Fig. 1-4.

There is a special prosthesis applicated in clinical, "Hydracadence knee: 1P50, (Proteor, France), which is a knee-ankle coupling designed transfemoral prosthesis, as shown in Fig. 1-5. The prosthetic knee joint is connected with ankle joint by a single-axis linkage mechanism, and the ankle joint is driven to rotate by the movement of the knee joint in gait. The knee damper is also hydraulically regulated. Compared with active prostheses, knee-ankle joint prostheses have the same advantages like other passive prostheses: simpler structure, lighter weight and cheaper price. At the same time, because of the special knee-ankle coupling designed characteristics, its gait performance is different from other passive prostheses. During walking, the prosthetic knee joint drives the ankle joint through

the connecting linkage rod, which makes the ankle joint dorsiflex or plantarflex actively. This feature is not available with other passive prostheses.

Fig. 1-5 Knee-ankle coupling designed prosthesis

Human knee joint is a double joint structure composed of tibial joint and patella joint, which will bear great joint force and moment in gait. Because of the double joint structure, the process of knee extension or flexion is complicated. The motion not only occurs in the sagittal plane but also in the coronal plane and transverse plane. The motion amplitude of joint rotation on sagittal plane is the largest. The performance test of a prosthesis is also comparing the motion characteristics of prosthetic knee joint to the human knee joint, and is carried out simultaneously in all three planes. The load on the knee joint of human body in gait is simulated in composite force.

1.3.3 Prosthetic ankle joint

A "prosthetic foot" consists of two parts, the ankle and the footplate of the prosthesis. These two parts are often designed into a whole. According to the characteristics of the ankle joint, it can be divided into the following categories (see

Fig. 1-6):

Solid Ankle Cushion Heel foot (or Static Ankle Feet). Abbrev. SACH foot. It is the simplest non-hinged prosthetic foot. SACH foot has a rubber wedge at the heel that cushions it from deformation when it hits the ground, reduces vibration and makes the lower limbs find a stable position on the ground. At the same time, different rubber heels can change the heel height of the prosthesis to cope with different choices. In the middle of the SACH foot there is a keel which is very hard and stable. This allows the prosthetic foot to provide good stability. The soft toe at the front of the prosthetic foot can deform at the end of the stance phase, compensate for the rotation of the metatarsal joint, and make the gait enter the swing phase smoothly. SACH foot also has drawbacks, such as that the heel cushion, while enhancing heel elasticity and reducing the impact of the heel landing, also delays the foot from initial contact to loading response. This allows the limb to prolong the stance phase to wait for the prosthetic foot to smoothly enter the next gait phase. The simple structure makes the SACH foot lighter, more durable, cheaper, and easier to maintain. These characteristics make it very popular, especially in some developing countries and regions.
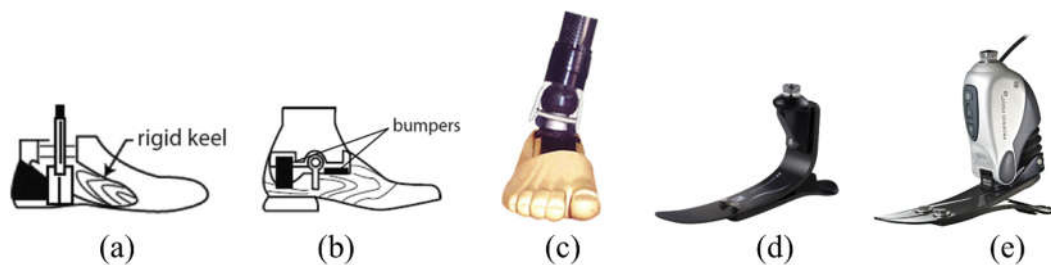
Fig. 1-6 Types of prosthetic feet: (a) SACH foot; (b) single-axis foot; (c) multi-axis foot; (d) energy-storing-and-releasing; (e) active foot prostheses (Versluys et al., 2009)

Single-Axis foot. Different from SACH foot, the ankle of a single-axis foot can rotate on the sagittal plane. It has an axis of rotation perpendicular to the sagittal plane for dorsal and plantar flexion of the prosthetic foot. Single-axial foot is designed with bumpers at the front and rear for cushioning while limiting and controlling ankle rotation and heel elasticity. The cushioning rubber block, acting in conjunction with the axis of rotation, allows the prosthetic foot to find a contact position faster than SACH foot in the double support phase, allowing the sole of the foot to be parallel to the ground faster and the center of gravity of the body to shift forward faster, thereby providing an early extension moment for the knee joint and improving the stability of the stance phase. Although the rotation range of the single-axis foot is smaller than that of the normal person, it can meet the requirement of walking on flat ground better than SACH foot. However, it is more difficult for such feet to walk on sloping roads (Zhao, 2005). Similar to the SACH foot, a single-axis prosthetic foot can also be replaced with a rubber block to change the heel height. In addition, the elasticity of the metatarsal joints also improves the stability of the amputee at the end of the stance phase.

Multi-Axis foot. As its name implies, its ankle can rotate in multiple planes (sagittal, coronal, and transverse) with greater freedom. Multiaxial prostheses are also elastic at the metatarsal joints and have similar characteristics to single-axis foot. Because of the special mechanism design, the multi-axis prosthetic foot can absorb certain ground reaction force on the transverse and coronal plane, and reduce the torque acting on the human body. This reduces the stress on the interface between the socket and residual limb, thereby reducing friction and skin irritation and providing better protection for the residual limb. Because of the multi-axis rotation, it can be better adapted to slopes or uneven road surfaces.

Energy-storing-and-releasing foot (ESR foot). The ESR foot can release the energy generated by compression during walking at an appropriate time so as to play an auxiliary pushing role. Because the amputation loses the ankle joint, the human body cannot actively plantarflex during the walk to generate enough propel force to push the person forward. The ESR foot can partially compensate for this loss. The ESR foot can store energy by elasticity deformation during the heel period, then release energy to push the prosthetic foot into the middle of the gait, and provide propulsion during the heel-off period. It is characterized by the storage of energy and the release of energy. This means that some of the energy consumed in the gait is stored in the prosthetic foot, making walking more efficient and easier for the amputee. These prostheses are designed in a variety of shapes, such as multi-axis and one-piece, etc. Most ESR feet are designed with separate toes at the toes to mimic the varus-valgus motion of the normal foot with greater stability. The front, middle and rear parts of the prosthetic foot are cushioned, and ankle joints are usually equipped with control and adjustment components. These designs play important roles in stabilizing gait, reducing energy consumption, absorbing vibration and improving stability. This kind of artificial foot satisfies the crowd mainly for the high physical activity user. South African disabled athlete Oscar Pistorius, for an example, won sixth place in the men's 400 m at the 2011 IAAF Challenge in Ostrava, Czech Republic, wearing an energy-storing-and-returning foot.

Active foot. Similar to active knee joints, such prostheses actively provide joint moment by means of an electric motor to cause dorsiflexion or plantar flexion of the joint (Armannsdottir et al., 2018; Jimenezfabian et al., 2017). It compensates for the loss of active moment-producing function of the ankle due to amputation.

At the same time, these prostheses cooperate with microsensor to detect gait characteristic parameters and adjust joint angle, joint moment or damping in order to adapt to more road conditions. These prostheses perform better on slopes and rough roads. Intelligent adjustment also makes the shoes wearable more abundant, can let the wearer choose running shoes, flat shoes, formal shoes and other different types. However, battery power, weight, maintenance, complex structure and high price have also seriously hampered the development of such prostheses, the prevalence rate is not high.

A passive prosthesis is combined with passive prosthetic knee and passive prosthetic ankle. One of the main drawbacks of passive transfemoral prosthesis is that the prosthetic limb is not able to generate enough propelling force during walking. The smaller anterior-posterior ground reaction force of amputated limb was presented in Schaarschmidt et al. (2012) and Cerqueira et al. (2013). That is one of the reasons why transfemoral amputees can not walk as fast as the able-bodied person could. The weakness in athletic ability also affects the movement of non-amputated limb.

1.4 Requirements for improving transfemoral amputees' gait

The methods for improving the gait of transfemoral amputees have always received substantial attention from researchers and prosthetists. Prostheses have a growing number of functions to meet different requirements of amputees, which enable amputees walk more comfortably and steadily. Nevertheless, there is still room for improvement. For instance, transfemoral amputees have demonstrated a higher metabolic energy cost than that of an abled person, and they need to utilize a specific movement for compensating the lost biomechanical function of the amputated limb

or the insufficient design of the prosthesis.

A more in-depth understanding of the gait deviation of transfemoral amputees is critical for the improvement of gait. The pathology of transfemoral amputees needs to be eliminated, and the burden of movement needs to be reduced. A further approach needs to be conducted to find the cause of gait deviation of transfemoral amputees. The prosthesis design can be improved accordingly and transfemoral amputees can lead a smoother and more coordinated gait.

1.5 Objectives of this study

Consequently, this study explores the gait deviation of transfemoral amputees by using biomechanical approaches and finds the potential to improve it. The objectives of this study are as follows:

To understand the gait deviation of transfemoral amputees and find the deficiencies of the current prostheses.

To explore the rationality and shortage of the knee–ankle coupling designed transfemoral prostheses.

To propose a new method for evaluating the amputees' gait or the performance of a prosthesis.

CHAPTER II LITERATURE REVIEW

Prostheses are designed to compensate amputees as much as possible for the loss of biomechanical function as a result of amputation. A good prosthesis should function as close as possible to the lower extremity of a human being. For this reason, scholars have conducted a series of evaluations through gait analysis to judge the merits and demerits of the prosthetic performance.

Owing to the advances in scientific research and the development of industrial technologies, the performance of transfemoral prostheses used in the clinic has been increasingly approaching that of abled persons. However, the prosthesis is far from being able to replace the lost musculoskeletal system in function, and there are many differences in gait between amputees and abled persons(Cerqueira et al., 2013; Chang et al., 2011; Highsmith et al., 2010; Schaarschmidt et al., 2012; Skinner and Effeney, 1985). The identification of gait deviation of transfemoral amputees and the evaluation of performance of a prosthesis are based on gait analysis.

2.1 Gait analysis basis

A gait is a periodic form and pattern in which people or animals move through their limbs. Gait analysis is the study of this motion system, or more precisely, generally refers to the study of human walking. Gait analysis refers to the process by which professionals record body movements through equipment measurements, and then systematically study body kinematics, kinetics, and muscle movements through observation and analysis (Whittle, 2003).

Gait analysis of transfemoral amputees is based on walking after wearing prostheses.

Through studying whether there is abnormal gait, the performance of prostheses and the degree of rehabilitation of amputees are judged. Gait analysis of prostheses is usually based on whether the angle, moment, and power of the joints (hip, knee, and ankle) in a gait cycle after wearing are close to those of abled subjects (control group).
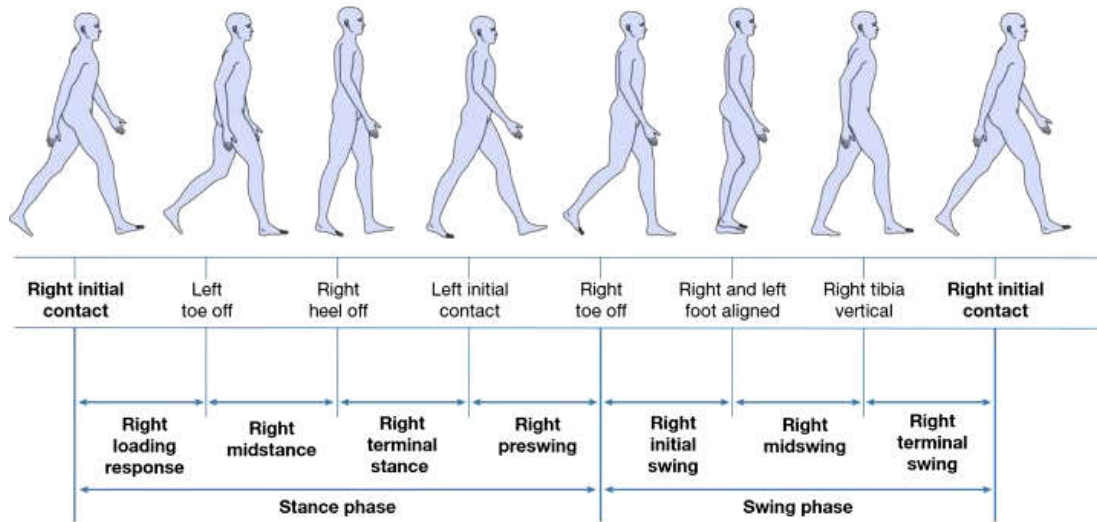
Fig. 2-1 Definition of the gait phase. (Picture adapted from JBJS Reviews)

Gait Cycle (GC) refers to the process that the heel of one side strike onto the ground to the next heel strike on the ground of the same lower limb during walking, as shown in Fig. 2-1. The periodic process of gait consists of a stance phase and a swing phase. The stance phase is the period of the foot contacting the ground, which accounts for 62% of the whole gait cycle of the abled persons. The swing phase is the period when the foot is off the ground, accounting for about 38% of the entire gait cycle of the abled persons. The stance phase is divided into two parts: double support phase (bipedal landing) and single support phase (only one limb landing). Both the ratio of the stance phase to the gait period and the ratio of the single support phase to the stance phase are close to the golden ratio (Iosa et al., 2013a). A

complete gait cycle consists of the following phase:

Initial contact: About 2% GC, the starting phase of the gait cycle. This is when the heel lands for the first time. Start with the first double support phase. At this point the lower limbs decelerate forward and the body finds a supporting position through the feet. If the action is abnormal in this period, the gait of the subsequent stance phase will be abnormal.

Loading response: Approximately 10% of GC, followed by full contact of the foot with the ground from the heel to the sole. The center of body mass begins to shift to the support foot.

Mid-stance: approximately 19% GC, during which the plantar surface comes into full contact with the ground. The contralateral lower extremity is in the swing phase, that is, the single support phase of the supporting foot. The center of gravity of the body rotates forward through the support limb.

Terminal stance: Approximately 19% GC, when the heel begins to leave the ground and ends with the opposite lower limb heel strike on the ground, i.e., the second double support phase begins.

Pre-swing: Approximately 12% of GC, from the time the opposite foot strike on the ground to the time the supporting foot leaves the ground.

Initial swing: Approximately 13% of GC, this period begins with foot support off the ground, knee flexion and move forward until reaches the maximum flexion angle.

Mid-swing: Approximately 12% GC, knee joint begins to extend until shank is

perpendicular to the ground.

Terminal swing: Approximately 13% of GC. Before the heel strike on the ground again, the lower extremity decelerates the movement and continues to move forward, prepares for the next heel strike on the ground.

Single support: Only one leg is supported in a gait cycle. The period from the opposite toe off the ground to the opposite heel strike on the ground (i.e., the opposite swing phase), when the ipsilateral lower extremity undergoes a mid-stance phase and a terminal stance phase.

Double support: The period during which both lower extremities are supported together during the gait cycle. There are two double support phases in the gait. The first one is the heel strike phase and the loading response phase, and the second one is the pre-swing period. The duration of double support is affected by walking speed, the faster the speed, the shorter the double support duration. If walking becomes running, the double support period disappears. The biggest difference between walking and running is whether there is a period of double limbs floating. If so, it is defined as running. The disappearance of double support is the turning point of walking and running, so it becomes the criterion of judging whether the walking race is against the rules or not.

The coordinate plane and direction of gait analysis are based on human anatomical position. When the human body stands, as shown in Fig. 2-2, it can be divided into three reference planes (Whittle, 2003): sagittal plane, frontal plane (or coronal plane), and transverse plane (or horizontal plane). The sagittal plane divides the human body into two parts: left and right; coronal plane divides the human body

into two parts: anterior and posterior; the transverse plane divides the human body into two parts: superior and inferior.
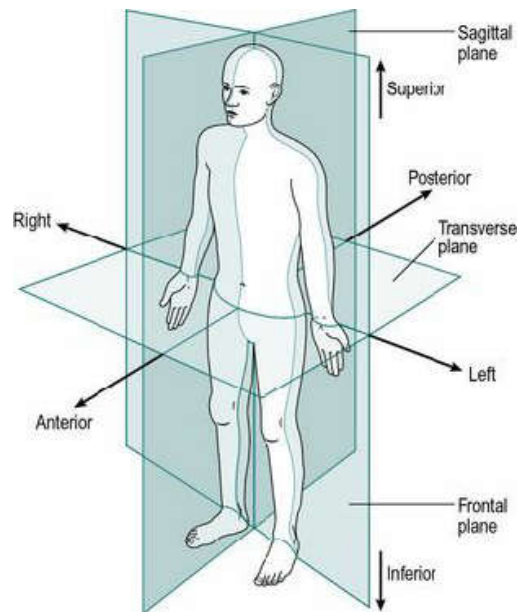
Fig. 2-2 The anatomical position definition in gait analysis (Whittle, 2003)

Direction definition: The umbilicus is anterior and the buttocks is posterior; The head is the superior end and the feet is inferior. The left and right are shown in the Fig. 2-2.

The medial side is near the centerline of the human body, and the lateral side is far away from the centerline of the human body. The proximal indicates proximity to the rest of the body and the distal end indicates the distance from the rest of the body. For example, the hip is the proximal part of the lower limb while the feet are the distal part of the lower limb.

Extension and flexion describe sagittal motion, such as hip and knee motion, as shown in Fig. 2-3. The sagittal movement of the ankle joint is generally referred to as plantarflexion and dorsiflexion, and is characterized by downward or upward movement of the foot (distal part) relative to the tibia (proximal part), respectively,

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as shown in Fig. 2-4.
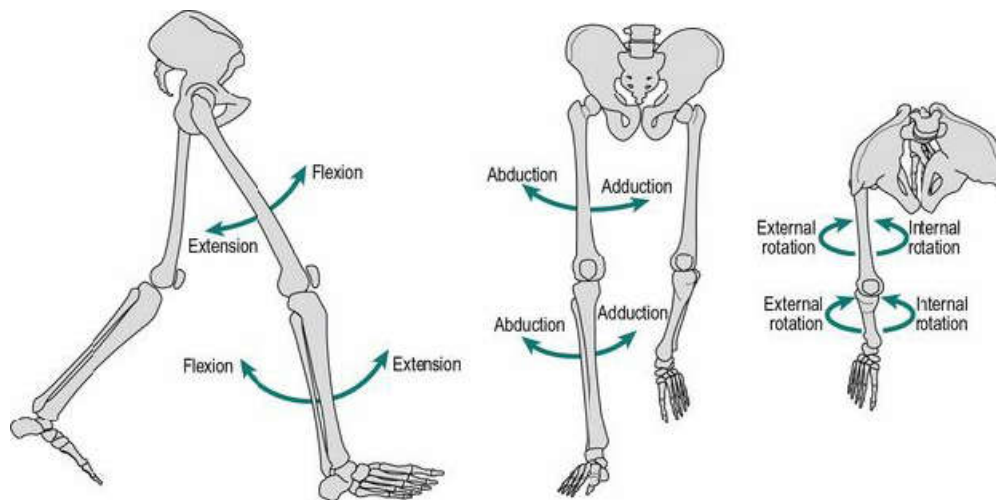
Fig. 2-3 Hip and knee motion direction definitions in gait analysis (Whittle, 2003)
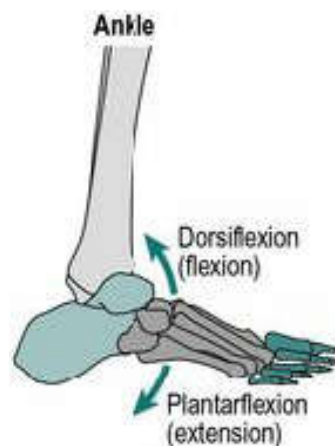
Fig. 2-4 Ankle motion direction definition in gait analysis (Whittle, 2003)

Abduction and adduction describe motion in the coronal plane. Where the movement of the distal part of the body away from and near the centerline of the body with respect to the proximal part, respectively.

Internal rotation and external rotation, which are also called medial and lateral rotation, describe the motion in the transverse plane. They represent the movement of the anterior surface of the distal segment relative to the proximal segment.

Generally, gait analysis consists of 16 basic gait parameters in 3D plane (Cimolin and Galli, 2014). There include three spatio-temporal parameters: the ratio of stance phase to gait duration, normalized velocity and cadence); and 13 kinematic parameters: average pelvic tilt, tilt amplitude, and mean pelvic rotation, minimum flexion of hip joint, flexion amplitude of hip joint, maximum abduction angle of hip in swing phase, mean rotation angle of stance phase, flexion angle of knee joint in initial contact, time of maximum flexion angle of knee joint, flexion amplitude of knee joint, maximum dorsiflexion of ankle joint in stance phase, maximum flexion angle of swing phase and mean forward angle of foot. However, these 16 spatio-temporal parameters and kinematic parameters are not enough to describe the gait difference of amputees, and kinetic parameters can be used as a powerful complement to the analysis of gait difference. Kinematics parameters can describe what happened very well, while kinetics can provide more analysis of the causes of what happened. These kinetics parameters are joint force, joint torque and joint power.

Gait analysis of transfemoral amputees is based on the above-mentioned basic gait parameters, most of which are aimed at hip, knee, and ankle joint angle, joint moment and joint power of healthy limbs and prostheses (Segal et al., 2006). At the same time, the ground reaction forces in three directions (vertical, anteroposterior and medial-lateral) are discussed (Cerqueira et al., 2013). In addition, temporal and spatial parameters are also a focus of gait discussion in transfemoral amputees (Highsmith et al., 2010). Depending on the purpose of the research, there is also an analysis of body trunk and pelvic movement (Goujon-Pillet et al., 2008), spinal stability (Corio, 2007), walking speed (Bonnet et al., 2010), etc.

Gait analysis is usually based on gait experiments with a motion capture system to obtain the above gait parameters. Motion capture systems are standard methods for obtaining gait kinematic and kinetic information by tracking luminous or reflective markers attached to the body in conjunction with force platform. This system usually requires special equipment and space. In addition, the emergence of wearable gait test equipment in recent years has made the system lighter and more applicable. Motion capture systems have been employed as an effective tool for health care, clinical diagnosis, and research.

2.2 Review of gait coordination study

Everyone's gait is unique. However, no matter what the distinctive characteristics of each gait, everyone can walk naturally, smoothly, harmoniously, and stably. Such movements can be unconscious and can be achieved without special effort. Gait coordination is often employed to describe the overall state of a gait in gait analysis. St-Onge *et al.* Believe that although human body movements are complex, the joints or parts of the human body are multi-degree of freedom when completing an activity task. However, the nervous system only needs comprehensive control to adjust the degree of freedom reaching the coordination (St-Onge and Feldman, 2003). This means that joint coordination can be ensured only by ensuring coordination in the degree of freedom. They speculate that there should be a basic coordination to ensure continuity of movement.

Gait coordination plays an important role in gait energy conservation and stability (Bruijn et al., 2011). It is defined as (Iosa et al., 2012): the ability to alternate, synchronize, balance, and rhythmically transfer body symmetry depending on coordination between limbs (Borghese et al., 1996), inter limbs (Reisman et al.,

2005), and between upper and lower limbs (Cappozzo, 1982; Iosa et al., 2007). Simultaneously, gait stability is defined as the ability to move parts of the body in a coordinated manner so that the body can walk at an appropriate speed and to minimize upper-limb oscillations (Cappozzo, 1982; Iosa et al., 2012).

Researchers study the coordination of gait through a variety of indicators. Iosa *et al.* made an assessment of gait coordination through the proportion of stance phases (Iosa et al., 2013b). They believe that coordination is to ensure smooth walking energy-saving features. Each gait cycle is a periodic repetition of a process. They analyzed the kinematics of two lower limbs over a period and found a golden ratio. As shown in Fig. 2-1, a gait period may be divided into two parts, a stance phase and a swing phase. The stance phase accounts for about 60%-62% of the gait cycle, while the swing phase accounts for 38%-40%. Such a ratio is considered an indicator of gait comfort and is used as an indicator of whether the gait is reasonable (Perry and Davids, 1992; Winter et al., 1990). Iosa *et al.* believe that the ratio of the stance phase to the swing phase should be the golden ratio. The golden ratio is not only reflected in the body proportion (Ferring and Pancherz, 2008), but also in aesthetics (Ricketts, 1982). Iosa *et al.* tried to use this theory to explain the coordination problem in gait and found that the gait stance phase of normal persons was about 61.7%, which was very close to the irrational number of golden section (0.618). At the same time, the swing phase is about 61.4% of the stance phase. In addition, the double support phase is about 58.4% of the stance phase. These ratios are all close to the golden ratio. Iosa *et al.* think that coordination is very important to gait and golden ratio can be employed as an important index of gait analysis.

Symmetry and limb dominance (also reflected as left-handedness or right-

handedness) are also indicators of gait coordination. Although gait symmetry can be defined in many ways, they all have in common the ability to describe the same behavior of both lower limbs. In order to simplify or build on some assumptions, the gait of an abled person can usually be considered symmetrical. The asymmetry of gait reflects the difference in lower extremity function. This difference is not a pathological one, but a reflection of the difference in propulsion and control tasks between the lower limbs (Sadeghi et al., 2000). Limb dominance is another explanation for the difference. To accomplish an operational task, humans are usually led by one leg and supported by the other. Therefore, the ground reaction forces exhibit asymmetry (Claeys, 1983). For amputees, stance phase of the prosthetic side is shorter and the ground reaction is lower after wearing the prosthesis (Skinner and Effeney, 1985a). Similarly, other pathological gaits, such as gaits with different foot lengths, can cause asymmetry. The study of symmetry can provide more information on the causes and effects of pathological gait. For example, muscle force is a good indicator of the balance of propulsion and control in a person's gait.

Muscle coordination is often employed to reflect patient's gait coordination. It is defined as the distribution of muscle activity or muscle force among muscles in order to produce a specified combination of joint moments (Prilutsky, 2000). The muscles of the lower limbs play an important role in gait. They produce muscle force to support the body against gravity and prevent the body from collapse; And generates a driving force to move the center of body mass forward; maintains balance on each plane in the gait (Perry, 1967). Pandy *et al.* studied the effect of medial and lateral muscle balance using a 3D model (Pandy et al., 2010). Gait experiments were carried out using image capture technology, EMG and reverse

dynamic calculation. The results show that the muscle force has a great influence on the acceleration of the center of body mass. Muscle force is important not only to support the body but also to produce propulsive force. On the coronal plane, muscle force also plays an important role in maintaining the balance of the body's center of body mass. Muscle forces coordinate the acceleration of the center of body mass on the coronal plane to keep the gait balanced and stable. Hwang and Abraham studied the muscle force of knee-ankle joint by EMG (Hwang and Abraham, 2001a, b). In their view, the central nervous system controls all parts of the body so that the whole can perform a motor task in a coordinated manner. They think that the movement of the knee muscles is related to the contraction of the ankle muscles. At the same time, the ipsilateral muscle movement is also related to the speed. This partly explains why neurological impairment leads to decreased muscular capacity and poor coordination in patients.

Joint coordination is also a common indicator used to analyze gait coordination. Continuous relative phase (CRP) is often employed for quantitative analysis of joint coordination. In this way, coordination is described and analyzed by continuous spatio-temporal parameters. CRP can be applied to study the phase angle difference between joints. Phase portrait can describe the position and motion vectors of joints or parts of the body during the gait cycle. Expression of joint angle and joint angular velocity on the phase portrait can effectively quantify the coordination between joints (refers to Section 3.3.1). CRP enables quantitative analysis of joint-to-joint interactions in a continuous manner. Chiu *et al.* analyzed gait coordination in 10 young and 10 old people. The results showed that velocity had a great influence on joint coordination. The effect is greater when the speed is lower (Chiu and Chou, 2012a). The variation of joint coordination in repeated gait cycle is also an

important indicator to verify the stability of gait. Shiu-Ling and Chiu used this variability to predict whether older people were at risk of falling. Because of the aging of the population, more and more attention has been paid to the prevention of falls for the elderly. According to Rubenstein, people over the age of 65 are at risk of falling about once a year, and falling is an important cause of death, but two-thirds of them are preventable (Rubenstein, 2006). In addition, principal components analysis (C. and H., 2017), vector coding (Hafer and Boyer, 2018; Yen et al., 2017) and other methods are often employed to analyze joint coordination.

There are various factors influencing gait coordination. For example, walking speed greatly influences on gait coordination, which is an important factor of gait. When walking speed is faster than the normal comfortable speed, the change of muscle coordination is not significant. When the speed is lower than the normal comfortable speed, the coordination will be greatly changed (Chiu and Chou, 2012b). Extreme value of hip extensor muscle force and knee extensor muscle force will increase with the increase of walking speed. But the change of plantarflexion muscle force of ankle joint is relatively small. During slow walking, the hip extensors and knee extensors contribute little to support the body and shift the body's center of gravity during the first half of the stance phase. This is because the task of supporting the body in this phase is performed primarily by the gluteus medius muscle, bone, and joints (Anderson and Pandy, 2003; Liu et al., 2008).

Not only walking speed but also age has an influence on the coordination of gait (Chiu and Chou, 2012a). Older people tend to walk at a slower speed to maintain a balanced gait (Winter et al., 1990). In addition, human walking can rely on the eyes to collect information, and then the central nervous system to regulate the various

parts of the body to achieve gait coordination (Patla et al., 1991). Trauma can also affect gait coordination. Shiu-Ling Chiu *et al.* found that shock had an impact on people's ability to control joint coordination. Hip-knee, knee-ankle coordination is critical in the event of an external impact on gait (Chiu et al., 2013). Other underlying pathologies, such as acoustics (Roerdink et al., 2007), may affect gait coordination. If the gait coordination is affected by these factors, the gait performance will be different from the normal. The walking speed, step frequency, stride length will be reduced. The stance phase is also more asymmetrical. Stroke sufferers face gait problems as described above. And acoustic pacing can help patients to recover their coordination. Open and close eyes also have a significant effect on the joint coordination of the supporting leg (Doherty et al., 2015)

In addition, some pathological problems have long-term effects on joint coordination (Doherty et al., 2014). For example, studies have found that amputations have an impact on joint coordination during movement (Mouchnino et al., 2006).

2.3 Conventional methods of gait analysis of transfemoral amputees

First of all, the research on transfemoral amputee gait is usually based on the analysis of the following parameters in gait: a) Spatio-temporal parameters (Sapin et al., 2006, 2008; SilverThorn et al., 2009; Torrealba et al., 2008; Versluys et al., 2009); b) Kinematic parameters (Cerqueira et al., 2013; Sapin et al., 2006, 2008; SilverThorn et al., 2009; Torrealba et al., 2008; Versluys et al., 2009); c) Kinect parameters (Castro et al., 2014; Sapin et al., 2006, 2008; SilverThorn et al., 2009;

Torrealba et al., 2008; Versluys et al., 2009); d) Gait symmetry (Schaarschmidt et al., 2012); e) Metabolic expenditure (Fey et al., 2012; Russell et al., 2017); f) Muscle force (Cerqueira et al., 2013; Ouellet, 2015; Pantall and Ewins, 2013); g) Muscle morphologic (Putz et al., 2017; Sherk et al., 2010), etc. The performance evaluation of the prosthesis is based on gait analysis of the above parameters. For instance, Segal et al. validated the performance of the knee joints of two different transfemoral prostheses, namely C-Leg (Otto Bock, Duderstadt, Germany) and Mauch-SNS (Ossur, Reykjavik, Iceland) (Segal et al., 2006). In the study, by comparing the spatio-temporal parameters (stride, walking speed) of the gait of the prosthesis that the amputee wears and the joint angle, joint moment and joint power of the hip joint, knee joint and ankle joint on the prosthetic side, the performance differences were compared, and the above parameters were compared with those of the abled persons, so as to judge whether the function of the prosthesis is reasonable.

However, the study of joint parameters mainly reflects the performance of a single joint, while the study on the motion of a single joint is less sufficient to reflect the motion status of the amputee. In case that the performance of a joint is improved while accompanied by the increase of the burden of other joints or even the lower limbs on the other side, the evaluation results for the single joint are not enough to show the motion state of amputees in an objective way. The musculoskeletal system of human body has a very high degree of freedom, and the control strategy of motion is extremely complex, while each part or joint of the body tends to always cooperate with each other in motion in order to ensure the accuracy and stability of motion. At the current stage, based on the commonly used methods for gait analysis of transfemoral prostheses, the coordination of amputees' gait can not be evaluated. Gait coordination is an important indicator concerning whether gait is stable or

energy economic (Bruijn et al., 2011), which is a key criterion used to measure a person's ability to move and to determine whether he or she can walk safely and accurately (Chiu et al., 2013; Chiu et al., 2010; Lu et al., 2008; Roerdink et al., 2007). Therefore, the gait deviation and the performance defects of transfemoral prostheses urgently need to be analyzed through the gait coordination research, in order to provide further theoretical guidance for clinical rehabilitation and the design of the prosthesis.

2.4 Knee-ankle coordination of transfemoral prosthesis

The traditional transfemoral prosthesis is knee-ankle split type, which means that the knee joint and ankle joint of the prosthesis are two relatively independent components. The knee and ankle joints of the prostheses for different amputees are likely to be the products of different brands, and the two prosthetic joints, the socket, and other linkage component are assembled and aligned by the orthotists to be a prosthesis, thus becoming the transfemoral prostheses worn by amputees. The knee and ankle joints of the split type transfemoral prosthesis work in an independent way and the two operate according to the design of mechanical structure respectively in gait, where the joints fail to collaborate with each other.

The musculoskeletal system of the human body involves much more complex motions. Moreover, the motion of lower limb joints has high degree of freedom, while the knee and ankle joints will be coupled to each other in order to achieve motion coordination. The motion of human body is usually a composite motion by multiple joints, and the joints interact with each other in a complex manner. Especially, because of the existence of bi-articular muscle, power can be transferred between joints, which enable the center of gravity of the body to be effectively

transferred according to the need through the rotation of body parts (Gj et al., 1987; Jacobs et al., 1996). The moment acting on any of the joints will not only cause the motion of the current joint but also cause the motion of other joints due to the coupling effect (Zajac and Gordon, 1989). Researchers call this moment, which affects other joints in multi-joint motion, as "interactive torque" (Dounskaia et al., 1998; Hollerbach and Flash, 1982; Naito and Maruyama, 2008). The interactive torque does not even depend on muscle contraction, but merely on the motion of adjacent joints.

A good prosthesis should have good walking ability. Pitkin believes that linkage of the knee and ankle joints of the prosthesis facilitates compliance and that this design allows the prosthesis to align itself in the gait, which is more beneficial to the amputee's gait (Pitkin, 1997).

Still, most of the products on the clinical application today are knee-ankle separated. Amputees can choose different prosthetic knee joints and prosthetic ankle joints according to their gait characteristics, as well as their own preferences and prices. The prosthetist combines the prosthetic knee joints and prosthetic ankle joints together, adjust the alignment and allows the amputees to wear. The knee or ankle of these prostheses work relatively independently in gait and cannot be adjusted in real-time according to the working condition of the other joint. Human walking is very complicated, which is a process of multi-dimensional interaction of various parts of the body. Body parts or joints interact and coordinate with each other in sagittal, coronal and transverse, which is three-dimensional space at the same time, so as to move the mass of center forward smoothly and stably in gait. The interaction between knee and ankle is a very important part of gait (Inman and

Eberhart, 1953). Sadeghi *et al.* Studied gait interaction between ankle and hip joints in normal subjects, but knee-ankle coordination still lacks sufficient attention (Sadeghi et al., 2001).

The coupling motion between the knee and ankle of the lower limb is of great significance to the gait. Studies based on healthy young adults overcoming obstacles have shown that coordination and stability of the knee-ankle motions contribute to maintaining the balance of human body (Lu et al., 2008). Chiu et al. also drew a similar conclusion through the gait and motions crossing obstacles in patients with concussion, and they believed that the knee-ankle coordination was of vital importance to the safety of foot lifting and forward motion (Chiu et al., 2013).

The knee-ankle interaction of the knee-ankle split type is much lower than that of the abled persons. Not only the prostheses of the traditional passive transfemoral prostheses are split type, but those of some "intelligent" (also known as "bionic") prostheses are also split type. Intelligent prostheses give feedback on the status of gait and external environment through sensors, and then the microprocessor adjusts the joints of the prosthesis in real-time, which will exert a positive effect on gait coordination, such as C-Leg (Otto Bock, Duderstadt, Germany). However, the adjustment of the chip is mainly focused on the knee joint, while the ankle joint is adjusted within a limited scope, and the ankle joint still maintains the passive feature of passive prostheses.

In recent years, the design of positive transfemoral prostheses has been proposed (Pillai et al., 2011; Sup et al., 2011; Sup et al., 2008). The positive transfemoral prostheses can systemically shift the knee and ankle joints and even the other parts of the prosthesis to the processor for unified adjustment, and the knee and ankle

joints are combined to cooperate with each other so as to achieve a coordinated state, but most of these designs are still in the research and development stage (see Figure 1-4 in Chapter I). In addition, due to the weight, maintenance and price of the product, it is less likely to publicize the positive transfemoral prostheses for the time being.

Among the passive transfemoral prostheses used in the clinic, there is knee-ankle coupling designed prosthesis, such as the prosthesis "Hydracadence knee：1P50" designed by Proteor (see Figure 1-5 in Chapter I). The biggest difference between the product and other passive transfemoral prosthesis is that the movement of the knee joint in gait can drive the rotation of the ankle joint, thus creating a joint linkage mechanism between the knee and ankle. The function of knee-ankle coupling designed is closer to that of human knee-ankle joint, which is what other passive transfemoral prostheses are lacking in. Sapin et al. analyzed the kinematics and dynamics of the mechanism through the traditional gait analysis method (Sapin et al., 2008), they believe that the linkage mechanism had certain advantages, especially its ability to increase the clearance height of toes off in the early swing phase. However, according to the results of Chapter IV, these parameters are still only targeted at the single joint, which has some limitations. Since it is a joint linkage function, it is not only necessary to analyze the performance of single joint, but also need to be validated from the perspective of the overall effects, such as joint-joint performance, to ensure whether the linkage function can play a positive role or induce a negative effect on the gait. Whether knee-ankle coupling designed design has the potential to improve gait coordination is worth further discussion.

2.5 Research on walking speed of transfemoral amputees

Compared with abled persons, transfemoral amputees tend to walk at a slower speed so as to ensure comfortable walking (Boonstra et al., 1993; Boonstra et al., 1996; Highsmith et al., 2010; Skinner and Effeney, 1985b). More than that, the maximum walking speed that amputees are able to achieve is also lower than that of abled persons (Genin et al., 2008; Russell et al., 2017).

Transfemoral amputees show a series of characteristics different from abled persons when they are walking on the flat ground at different speeds. For instance, the stance phase of amputated limb is longer at a lower walking speed, and the knee joint of the non-amputated limb has a higher loading rate at a fast walking speed (Bonnet et al., 2010; Esposito et al., 2015; Schaarschmidt et al., 2012). With the increase of walking speed, unilateral transfemoral amputees show a number of changes in gait, such as the increase in the asymmetry of the ground reaction force of bilateral lower limbs in the vertical direction, and the decrease in the asymmetry in the swing phase (Nolan et al., 2003). Nolan et al. believed that the gait frequency would increase in order to raise the walking speed. However, the asymmetry of the time parameters of the transfemoral amputees' bipedal gait, such as the time proportion of swing phase, which will affect the increase of step frequency and thus restrict the increase of walking speed (Nolan et al., 2003). Decreased control of the walking speed shows some defects in the motion ability of transfemoral amputees.

Walking speed also has a great influence on the metabolic expenditure of amputees. The amputees had higher metabolic expenditure than abled persons during walking (Detrembleur et al., 2005; Waters and Mulroy, 1999). According to the research data of Russell et al., the metabolic expenditure of transfemoral amputees during

walking was about 46% higher than that of abled persons on average (Russell et al., 2017). From the research on metabolic expenditure of unilateral transfemoral amputees in walking, Detrembleur et al. found that metabolic expenditure per gait cycle at the low walking speed was twice as high as that at the preferred speed they feel comfortable (Detrembleur et al., 2005). Genin et al. studied the metabolic expenditure of unilateral transmural amputees walking within the specified distance and found that the relationship between walking speed and metabolic expenditure of the amputees was U-shaped and that there was an optimal walking speed (Genin et al., 2008). The increase of metabolic expenditure is another important factor affecting the walking speed of amputees (Russell Esposito et al., 2015).

Previous studies on the gait speed of transfemoral amputees mainly focused on the analysis of spatio-temporal parameters, kinematic parameters, and dynamics parameters, as well as the influence of different walking speeds on metabolic expenditure, while neglecting the impact of speed on gait coordination. The adaptability of gait coordination under various speeds is the embodiment of athletic ability. Inter-joint coordination is an important indicator of gait coordination, which is of great significance to the dynamic balance in walking (Lacquaniti et al., 1997; Yen et al., 2017). Chiu and Chou pointed out that the change of inter-joint coordination of the gait in the stance phase is extremely important for the stability of gait (Chiu and Chou, 2012b). The inadequacy of regulating ability in inter-joint coordination will lead to gait deviation (D A et al., 1990). With the change of walking speed, the inter-joint coordination of lower limbs will change accordingly (Chiu and Chou, 2012b). The weakening of control over different walking speeds may exert an effect on the daily life of transfemoral amputees. Up to now, how walking speed affects the inter-joint coordination of the lower limbs of transfemoral

amputees remains to be unknown. The lack of understanding of this impact may directly hinder the improvement of prosthetic design and clinical rehabilitation.

2.6 Review of the rehabilitation of non-amputated limbs for transfemoral amputees

After wearing the prosthesis, transfemoral amputees are still unable to achieve the same ideal gait as abled persons. In order to achieve a more stable and harmonious gait, the lost biomechanical function will be partially made up for by the non-amputated limb through a compensatory mechanism (Schaarschmidt et al., 2012). The non-amputated limb of the unilateral amputee has a sound musculoskeletal system and biomechanical function, but in order to coordinate the changes in the motion pattern of the prosthetic side, the kinematic and dynamic parameters of the non-amputated limb will also change partially compared with the abled persons (Castro et al., 2014; Russell et al., 2017; Russell Esposito et al., 2015). In the daily life of amputees after going into rehabilitation and wearing the prosthesis, not only the residual limb has more pathological problems, but also the non-amputated limb has a high incidence of joint pain (Devan et al., 2012; Hagberg and Brånemark, 2001) and osteoarthritis (Kulkarni et al., 1998; Norvell et al., 2005).

Joint pain or osteoarthritis is often caused by the increase of long-term loading on the joints (E Hurwitz et al., 2001), while the stress of the joint of the amputee's non-amputated limb in gait is different from that of abled persons in many aspects. Although the vertical force given to the amputee's non-amputated limb by the ground is similar to that of abled persons, the anterior-posterior force given by the ground is less than that of abled persons (Castro et al., 2014; Cerqueira et al., 2013).

Influenced by the reaction force of the ground, the maximum loading rate of non-amputated limb under the reaction force of the ground is larger than that of abled persons (Esposito et al., 2015), while the knee adduction moment is larger, but the hip adduction moment is smaller (Chang et al., 2011; Segal et al., 2006). Chang et al. suspected that the pain of the joint in the amputee's non-amputated limb was bound up with the joint moment suffered by the joint in the long term (Chang et al., 2011).

Joint moment is an important parameter that reflects the motion status of the joint. However, if the muscle force were not studied, the joint moment would be insufficient to reflect the internal motion of the joint. The function of muscles in gait is to generate the muscle force so as to produce joint moment and drive or control the motions of the joints (Levine et al., 2012). When discussing the moments of joints, the same joint resultant moment can be obtained by varying degrees of contraction of each muscle (Johansson et al., 2005). Inoue et al. discussed the action of them on the joints through the motion of antagonistic muscles (Inoue et al., 2011), believing that the joint moment of a joint may remain unchanged and the stiffness of the joint will increase due to the simultaneous action of flexor and extensor muscles of the joint. Thus, it can be seen that joint moments and muscle force are of equal importance in assessing the performance of joints in gait and exploring the compensatory gait, and can not be dispensed with.

The improvement of performance of the prosthesis should not be at the expense of the health of the non-amputated limb, otherwise the performance improvement would go just the opposite. Moreover, the goal to improve the performance of the prosthesis should not be to make the joints of the prosthesis perform closer to that

of abled persons but to minimize the impact on the non-amputated limb and parts of the body after the amputee wears the prosthesis. Even though the mechanical joints of the prosthesis operate in a totally different way from those of the abled persons, it is an ideal prosthesis only if it can provide sufficient compensation for the biomechanical function. The less compensatory mechanisms are needed for the non-amputated limb or other joints and parts of the body, the more ideal the prosthesis will be. The muscle coordination of the non-amputated limb is another manifestation of whether the gait is healthy and reasonable after the amputee wears the prosthesis (Zajac et al., 2003).

Muscle coordination was defined (Prilutsky, 2000) as the distribution of muscle activation or muscle force required to achieve a specific resultant moment of the joint. The studies of the muscle force of the transfemoral amputees usually focus on the residual limb. Putz et al. (2017), Jaegers et al. (1995), Pantall and Ewins (2013) all studied the muscle of the residual limb or that of the hip joint of the residual limb, analyzed the changes of muscle morphology and explored the muscle atrophy of the residual limb. Although Wentink's study of the muscle force of the amputee's residual limb also analyzed four calf muscles of non-amputated limbs (Wentink et al., 2013), while Cerqueria et al. discussed forces of four calf muscles of non-amputated limbs in the research to study the ground reaction forces of the transfemoral amputees (Cerqueira et al., 2013), but they are limited to the measurement of muscle force. In this regard, there are only a small number of studies in this aspect, which makes it difficult to check the overall muscle coordination of the amputee's non-amputated limb. Up to now, there is limited research on the muscle coordination of the non-amputated limbs of unilateral transfemoral amputees.

One of the important reasons for the lack of research on muscle coordination of the non-amputated limb is the limitation of research techniques. The muscle force of lower limbs is often measured by electromyography (EMG) technology on muscle activity, and then muscle force is analyzed. Wentink et al. found that the muscles of the non-amputated limb increased the activation duration so as to prolong the stance phase, and provide more propulsive forces to propel the body forward, while the muscles of the residual limb are all at an active state at the end of the stance phase and in the swing phase, in order to ensure the fixation of the socket and the stability of the gait (Wentink et al., 2013). This finding was supported by Cerqueira et al., while it was also found from their research that the activation of muscle strength of the non-amputated limb of transfemoral amputee lasted longer (Cerqueira et al., 2013). Ouellet studied the muscle activity of transfemoral amputee's bilateral muscle through five daily activities tasks and found that the EMG signal of the muscle of the residual limb was 1/3 weaker than that of the muscle of the non-amputated limb (Ouellet, 2015). However, estimating muscle force from EMG remains challenging with a variety of assumptions and signal conditioning/filter techniques. Muscle force obtained from EMG signals is influenced by the variety of human factors, such as signal conditioning or filtering methods. Wentink et al. doubted the accuracy of the high variability of EMG data (Wentink et al., 2013), although Wong et al. suggested that the variability could be caused by a higher vulnerability to muscle fatigue (Wong et al., 2015). It has been proved by facts that the anatomy and location of the muscle may change after the amputation surgery, which affected the validity and accuracy of the placement of EMG electrodes (Wentink et al., 2013). In addition, the calculated muscle force and joint moments are less likely to enforce the equilibrium equation. There are still some other

limitations in the EMG technique. For deep muscles or muscles with small muscle bundles, it is difficult to measure them with the EMG technique, such as posterior tibial muscles and popliteal muscles. It is also one of the reasons why the research on the muscle force of the amputee's non-amputated limb fails to be carried out in an all-round way.

An alternative solution is to use the inverse dynamics analysis of the musculoskeletal model to obtain the forces of various muscles (Bae et al., 2008). Inverse dynamics analysis can make up for the shortcomings of EMG technology as mentioned above, so that we will be able to analyze the muscle coordination of lower limbs in a quick and comprehensive way. Inverse dynamics is a method to calculate the dynamic forces dynamic force that changes the motion of an object by finding a solution through the laws of mechanics according to the presentation of its motion (Shi and Wei, 2003). The concept of inverse dynamics is relative to direct dynamics. Direct dynamics is to solve the motion state and force status based on the given conditions such as external force and moment, while inverse dynamics, on the contrary, is to calculate the driving force and moment needed to produce this state based on the given motion state. In terms of gait analysis, inverse dynamics analysis is used to calculate the forces on the joints, bones, muscles, and muscle tendons of the body based on the kinematic parameters measured in the gait. Through inverse dynamics analysis, not only parameters such as joint force and muscle force can be obtained, but also deformation of skeleton or muscle, elastic energy storage of tendon, and analysis of the role of antagonistic muscle and other data can be concluded. Inverse dynamics analysis is based on the musculoskeletal system model of the human body. In order to obtain more accurate results, researchers have optimized the musculoskeletal system in various aspects in recent

years (Klein Horsman et al., 2007; Manders et al., 2008; Petrella et al., 2013; Weber et al., 2015). Among them, the lower limb model of MoCap Model (AnyBody Technology A/S, AMMR 1.6.3) has been optimized with better achievements and applied in a variety of research fields (Klein Horsman et al., 2007; Manders et al., 2008; Petrella et al., 2013; Weber et al., 2015).

2.7 Research gap in previous studies

The studies motioned above lack the analysis of gait coordination of a transfemoral amputee, and the characteristics of gait coordination of amputees are still unknown. The lack of understanding of amputees' gait coordination slows the prosthetic improvement and clinical rehabilitation process.

The knee–ankle coupling designed mechanism of a transfemoral prosthesis is closer to the knee–ankle joint movement mode of the musculoskeletal system of human beings, which enables the knee and the ankle to cooperate with each other in the gait. It is exactly what other passive transfemoral prostheses lack, but it is urgently needed to validate whether such linkage function can effectively improve the gait coordination of amputees.

The walking speed is an important manifestation of athletic ability. It is inevitable for human beings to adopt different walking speeds in daily walking so as to cope with the external environment and achieve different athletic purposes. However, the transfemoral amputee's ability to control the walking speed is far lower than that of abled persons. In this regard, it is urgent to clarify the different characteristics of gait coordination of amputees walking at varying speeds.

The state of muscle coordination of a non-amputated limb in the gait has not been

established. In previous studies, the non-amputated limb has been studied more in comparison to the prosthetic side, and less attention has been paid to the non-amputated limb itself. The performance of a prosthesis should not be improved at the expense of the health of the non-amputated limb. Studies have shown that there is a high incidence of pathological problems in the non-amputated limb, such as knee arthritis. There is still a lack of explanation for the etiology. Thus, we need to know whether the non-amputated limb is adversely affected by the prosthetic gait. Moreover, the muscle coordination of a non-amputated limb is another manifestation of gait coordination, which enables us to better understand the characteristics of amputees' motor control.

2.8 Research scope of this study
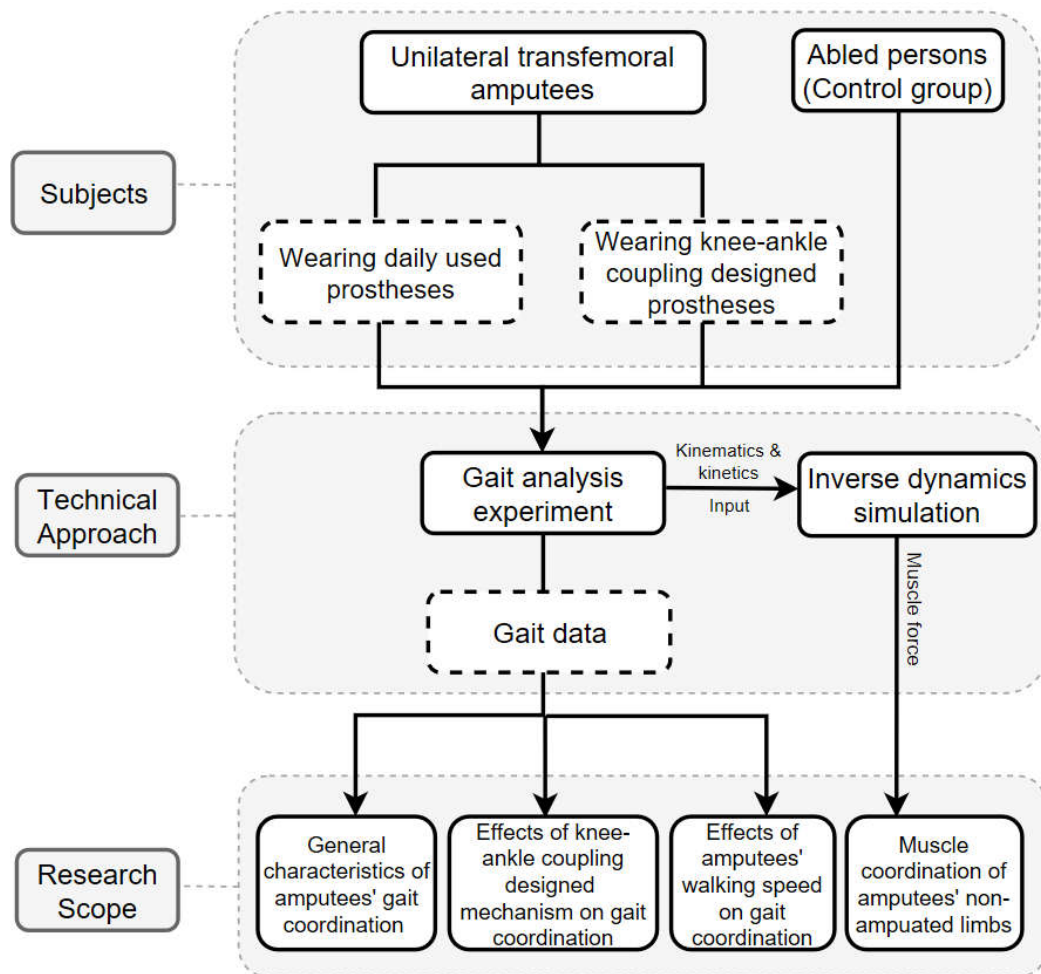
Fig. 2-5 Outline of this study

This study was carried out in accordance with the outline shown in Fig. 2-5. This work regarded unilateral transfemoral amputees as the research subjects to study the characteristics of gait coordination, and then conducted a comparative analysis with abled persons treated as the control group. Technical means include gait analysis and inverse dynamics simulation. The gait analysis is used to capture the kinematic, kinetics, and space–time parameters of gait, and the inverse dynamics analysis is used to estimate the muscle forces of lower limbs in gait. This paper is divided into four parts, as follows:

The first part focuses on the prosthetic gait with traditional passive transfemoral prostheses, which is the most widely used prostheses in the clinic. By analyzing the inter-joint coordination and decomposition index (DI) of the amputees' gait at their preferred walking speed (the speed at which they feel comfortable), the general features of amputees' gait coordination were obtained by comparing with those of abled persons.

In the second part, we compared gait coordination of amputees wearing the knee–ankle coupling designed prostheses with that when wearing the traditional passive transfemoral prostheses, and it was validated whether the knee–ankle coupling designed function can have a positive impact on gait coordination.

In the third part, gait analysis experiment on amputees were carried out with three speeds: their preferred speed, fast speed, and slow speed. The differences in gait coordination among the different speeds were analyzed, and the connections between coordination changes and motion control were revealed.

In the fourth part, inverse dynamics were implemented with subject-specific musculoskeletal models to estimate muscle forces during gait. Non-amputated limb muscle coordination of unilateral transfemoral amputees was presented. Muscle coordination realignment of amputees compared to abled persons was discussed, and its implication for rehabilitation and prosthesis design was proposed.

CHAPTER III RESEARCH METHODOLOGY

The methodology of this study mainly included gait analysis and inverse dynamics simulation. The gait analysis was conducted to obtain various parameters including gait duration, walking speed, stride length, joint angle, joint moment, and ground reaction force (GRF). The inverse dynamics analysis was used to calculate the muscle forces of lower limbs during walking.

3.1 Gait analysis experiment

Fig. 3-1 Demonstration of gait analysis

This project has been approved by the Human Subject Ethics Sub-committee of The Hong Kong Polytechnic University. The reference number is HSEARS20170117001. The experiment was conducted in "Department of Biomedical Engineering of The Hong Kong Polytechnic University" and "National Research Center for Rehabilitation Technical Aids of Ministry of Civil Affairs of the People's Republic of China". The content of the experiment was completely

informed to the subjects before the experiment through the "Information Sheet" and the "Consent to Participate in Research" was signed after obtaining the consent from the subjects.

3.1.1 Gait analysis experiment equipment

a) 3D Motion Capture System (Vicon, Oxford Metrics Ltd., Oxford, United Kingdom), eight infrared cameras were included.

b) Two AMTI dynamometers (OR6, AMTI, Watertown, United States).

c) Processing software for experimental data: Nexus 2.5.

3.1.2 Subjects recruitment

1) Eight unilateral transfemoral amputees;

2) Eight healthy adults.

Inclusion criteria:

i. Inclusion Criteria for amputees:

a) Unilateral transfemoral amputation;

b) Aged from 18-60 years old, no limit in gender;

c) Have received rehabilitation training and worn a passive prosthesis for more than half a year, has fully adapted to the prosthesis;

d) Able to read "Information Sheet" and sign the "Consent to Participate

in Research";

e) Independently walker.

Ii. Exclusion criteria for amputees:

a) Amputees with communication disorders such as language disorder or dysgnosia;

b) There are other pathological diseases that affect gait, such as: muscle or nerve diseases, other musculoskeletal surgeries besides amputation, ulceration or pressure sores in residual skin or tissues, etc.

Iii. Inclusion criteria for healthy adults:

a) Aged from 18-60 years old, no limit in gender;

b) With normal walking ability;

c) Able to read "Information Sheet" and sign the "Consent to Participate in Research";

Iv. Exclusion criteria for healthy adults:

a) Subjects with communication disorders such as language disorder and dysgnosia;

b) There are other pathological diseases that affect gait, such as: muscular neuropathy, length feet or musculoskeletal injuries.

Table 3-1 Subject characteristics

Control group				Transfemoral amputees			
Subject NO.	Age (years)	Weight (kg)	Height (cm)	Subject NO.	Age (years)	Weight (Kg)	Height (cm)
1	22	58	178	1	23	60.3	168
2	22	55	171	2	30	54.4	174
3	27	82	181	3	24	69.5	188
4	21	63	173	4	27	66.3	169
5	36	63	183	5	48	82	185
6	23	57	174	6	32	80	172
7	25	73.3	173	7	32	71.8	170
8	32	62.1	171	8	27	78.1	175

The information of the subjects finally recruited is shown in Table 3-1 and Table 3-2. All subjects were male. Amputee age: 30.4 ± 7.9 years old; height: 175.1 ± 7.5 cm; body weight: 70.3 ± 9.7 kg. Healthy adults (control group) age: 26 ± 5.4 years old; height: 175.5 ± 4.6 cm; body weight: 64.2 ± 9.1 kg.

3.1.3 Experimental procedure

Gait analysis A: This experiment was conducted prior to Gait analysis B. The amputees were required to wear their daily used prostheses to participate in the experiment. The same experiment was performed in control group. A demonstration of the experiment is shown in Fig. 3-1. The experiment was conducted as the following procedure:

1) Subject preparation: Subjects wore tights clothes, shorts, and flat shoes. Thirty-nine retroreflective markers were affixed on the surface of the human body according to Plug-In Gait Model (Kim et al., 2014), which is shown as Fig. 3-2. After the completion of the above preparations, a warm-up exercise of 5 minutes of slow walking on the treadmill was performed.

2) Data acquisition procedure: Subjects were instructed to walk along a 10-m straight level walking path under the capture range of the Vicon system. Kinematic data was captured by 8 infrared cameras. Synchronized kinetic data was acquired by two in-ground force platforms. A gait cycle started with one foot contacting the ground ('initial contact') and ended with the same foot contacting the ground again (Versluys et al., 2009). The participates were instructed to trace a straight line while avoid targeting the force platform when walking over it. A successful trial was considered when (1) one in which the foot landed completely on the force platform during the stance phase; (2) there was no attempted alteration in the walking style as judged subjectively by the investigator. Two continuous gait cycle of both left and right lower extremities were involved of each trial. Three sets of successive trial data were collected for each gait test content (such as different walking speed). Considering that random error may occur and some coordination features may be cross-out if we try to average the data of the three trials, only one trial dataset was randomly selected for subsequent analysis.

3) Software settings: Kinematic and kinetic data were sampled at 200 Hz and 2000 Hz, respectively. Data processing was conducted using software Nexus (Vicon, Oxford Metrics Ltd.), and were filtered using a fourth-order, Butterworth, low-pass filter at 6 Hz and 120 Hz., respectively (Chiu et al., 2015; Hutin et al., 2012).
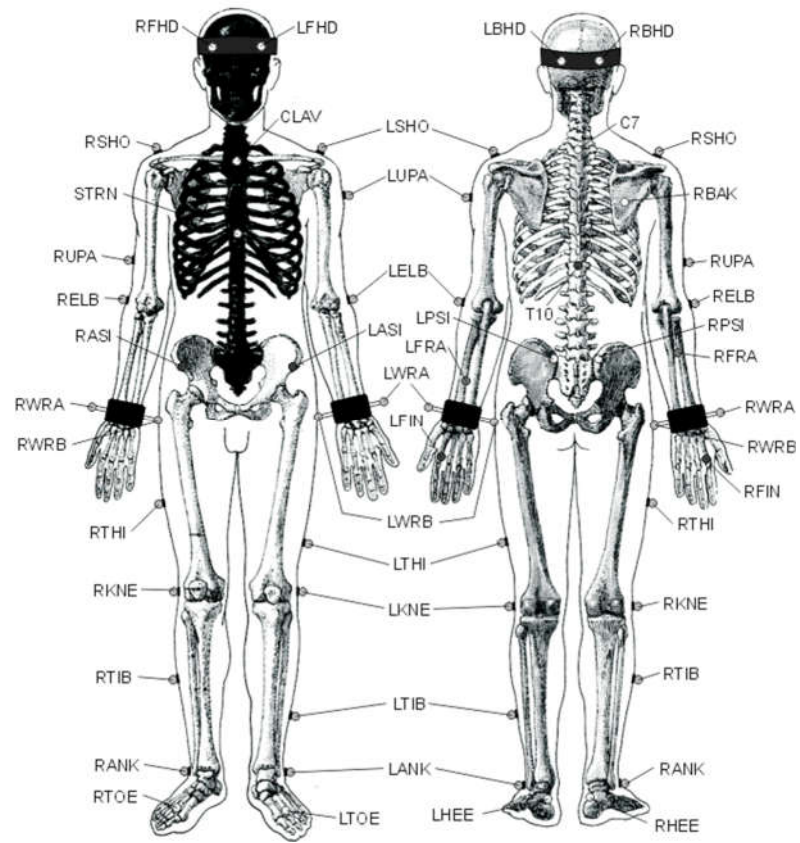
Fig. 3-2 Plug-in Gait Model markers setting (Vicon®, 2010)

Table 3-2 Prosthetic data of amputees

Subject NO.	No. of years since amputation	No. of years using current prosthesis	Amputated side	Prosthetic knee	Knee Damping	Prosthetic foot	Foot Type
1	13	4	L	Jingbo JB-850	Pneumatic	Jingbo FY-YDJ	ESR
2	1.5	1.5	R	Ottobock 3R80	hydraulic	Ottobock 1A30	Mult-Axis
3	18	10	R	Jingbo JB-850	Pneumatic	Jingbo FY-YDJ	ESR
4	5	0.5	L	Ottobock 3R80	hydraulic	Ottobock 1S101	SACH
5	17	6	L	Ottobock 3R80	hydraulic	Ottobock 1C40	ESR
6	7	5	R	Jingbo JB-951	Pneumatic	Jingbo FY-YDJ	ESR
7	15	1	L	Jingbo JB-810	Pneumatic	Jingbo FY-YDJ	ESR
8	5	5	R	Nabtesco NK-6	hydraulic	Jingbo FPJ-4	SACH

L/R:Left/Right。 ESR: Energy Storing and Releasing prosthetic foot; SACH: Solid Ankle Cushion Heel foot。

Gait analysis B: The subjects wore knee-ankle coupling designed prostheses (as shown in Figure 1-5) for gait analysis. The procedure is as follows:

1) Prosthesis preparation: The anthropometric of each amputee was measured when conducting Gait analysis A. Appropriate subjects among the 8 amputees were selected for wearing the knee-ankle coupling designed prosthesis since there is a requirement for the length of prosthetic shank. And the prosthesis is relatively heavy, a strong body is needed for a good adaptation. The knee-ankle coupling designed prostheses were prepared in advance by the prosthetist, as shown in Fig. 3-3. The prosthetist removed the prosthetic sockets from the daily used prostheses to the knee-ankle coupling designed prostheses and adjusted the alignment. When the prosthetist adjusted the realignment, the length of the prosthetic limb was adjusted to have the same length of the non-amputated limb. The location of the rotation centers of the two prosthetic joints may be a little different due to the different brands. No more procedure was adapted to adjust the location of the rotation centers except for ensuring equal length of the both lower limbs. The original alignment was marked before removing the prosthetic socket to ensure that the original alignment will not be changed after reassemble again;

2) Subject preparation: Amputees wore the knee-ankle coupling designed prostheses under the instruction of prosthetist until they can walk with it fluently and safely, this was judged by prosthetist. More than half a day was needed before the gait analysis for allowing amputees have a good adaption to the knee-ankle coupling designed prostheses. Gait analysis would not be conducted until receiving the permission of both the prosthetist and amputees.

3) The gait analysis process is the same as Gait Analysis A;

4) After the experiment, the prosthetic sockets were re-fixed to the daily used prostheses of the amputees, and the amputees wore them. (The alignment of the amputees' original prostheses will not be changed after the process).

The prosthetist participated in all the experiments for ensuring the safety of the amputees.

As the main movement of the lower extremity joints during the gait occurs in the sagittal plane, so the study of the inter-joint coordination on the sagittal plane is very important. Therefore, the study of gait in this paper focuses on the sagittal plane only. In addition, the gait parameters studied in this study have been proved to have no significant difference between the two lower limbs of normal subjects. Therefore, the data of lower limb of normal subjects in the study were averaged bilaterally (Chiu et al., 2015; Chiu and Chou, 2012b; Chiu et al., 2010).

Fig. 3-3 Prosthetist prepares the knee-ankle coupling designed prosthesis for the amputees

3.2 Inverse dynamics analysis

In this study, information including kinematics and kinetic parameters collected in gait analysis will be used as input for inverse dynamics analysis.

<center>(a)</center>   <center>(b)</center>
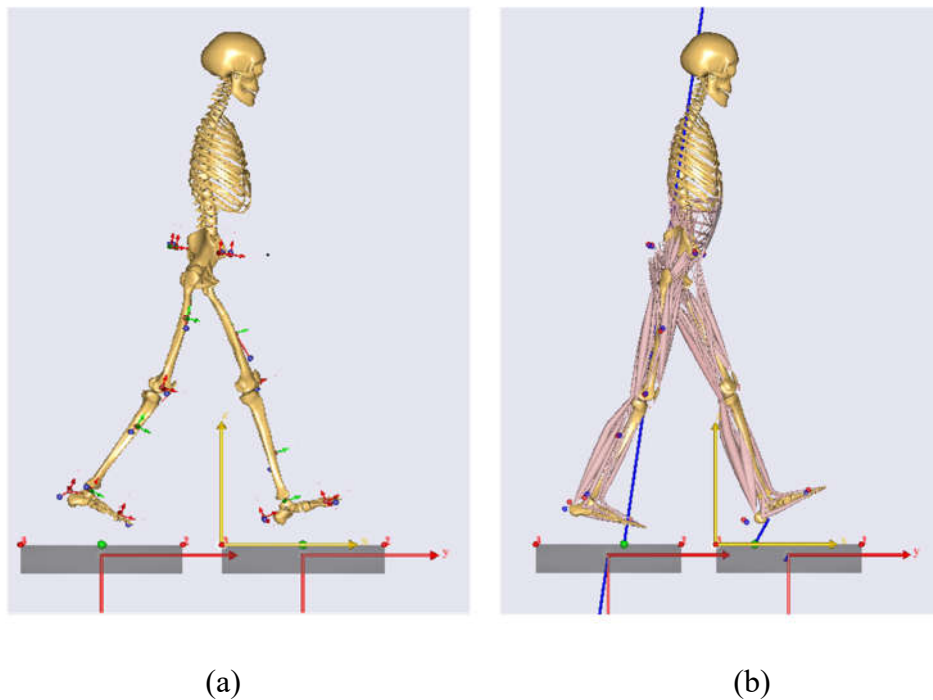
Fig. 3-4 Graphical representation of the inverse dynamics analysis process: (a) kinematics fit; (b) muscle force calculation

Muscle recruitment in inverse dynamics is the process of determining the set of muscle forces that will balance a given external load in the rigid-body dynamical system. The redundancy in the solutions is optimized by different control solutions. Different optimization methods for the redundancy in the muscle recruitment system and different musculoskeletal models might result in different estimates for muscle forces.

Inverse dynamics analysis was performed using the commercial software Anybody (AnyBody Technology A/S, Aalborg, Denmark). The general human

musculoskeletal model, MoCap Model lower extremity model (AMMR 1.6.3), which has been verified was used in a musculoskeletal model (Klein Horsman et al., 2007; Manders et al., 2008; Petrella et al., 2013; Weber et al., 2015). MoCap Model lower limb muscle model includes a total of 37 bundles and 159 bands. The calculation of muscle force is based on the muscle recruitment criteria defined by the Anybody software, and the balance equations of force and moment were established for the muscles involved in the calculation in the model, thereby the muscle force at each moment was obtained. The maximum value of muscle force is related to the height, weight, and length of the body segment of the subject, and is also related to the scaling of the body parts. Therefore, parameters of the subjects are specified according to anthropometry (Weber et al., 2015). It is worth mentioning that, considering the difference between the weight of the prosthesis and the lower limbs of the normal person, the ratio of the weight of prosthesis in the body weight is adjusted according to the measurement to improve the calculation accuracy (Dabiri et al., 2009; Hekmatfard et al., 2013). The inverse dynamics analysis is shown as Fig. 3-4.

The musculoskeletal system used in inverse dynamics analysis simplifies the human body into rigid body model. Taking the elbow joint as an example (as shown in Fig. 3-5), if the magnitude of the external force and the loading position, the size of the forearm and the attachment point of the biceps to the bone are known, then the muscle force and joint force can be calculated based on the equilibrium equation of the space force system (as in equation 3-1).
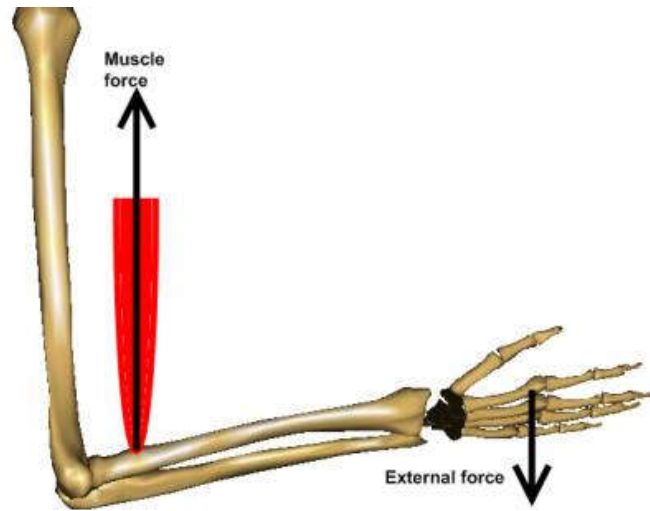
Fig. 3-5 Example of inverse dynamics analysis of the elbow joint musculoskeletal model (AnyBody Technology, 2014)

The Cartesian coordinate system was established with the simplified center O as the origin of the coordinate, and the equilibrium equation of the spatial arbitrary force system is shown in equation (3-1).

$$\left.\begin{array}{l} \sum F_{ix} = 0 \\ \sum F_{iy} = 0 \\ \sum F_{iz} = 0 \\ \sum M_x(F_i) = 0 \\ \sum M_y(F_i) = 0 \\ \sum M_z(F_i) = 0 \end{array}\right\} \qquad (3\text{-}1)$$

The musculoskeletal system produces multiple degrees of freedom of movement in gait, including the joint action of multiple joints and multiple muscles. The same joint will have a joint interaction of multiple muscles, even the interaction of agonist and antagonist. In addition, there are muscles that span multiple joints, thus the muscle contraction affects multiple joints at the same time. The redundancy in the muscle recruitment system was optimized using the muscle recruitment methods

integrated in the Anybody Modeling System as shown in the Equation (3-2), (3-3), (3-4) (Skals et al., 2017). These methods solve a polynomial optimization problem that minimizes a cost function G, subject to the dynamic equilibrium equations and to non-negativity constraints ensuring that the muscles can only pull but not push and that the muscle forces remain below the strength.

$$\min_{f} G(f^M) = \sum_{i=1}^{n^{(M)}} A_i \left(\frac{f_i^{(M)}}{N_i} \right)^3 \tag{3-2}$$

$$Cf = r \tag{3-3}$$

$$0 \leq f_i^{(M)} \leq N_i \qquad i = 1, ..., n^{(M)}. \tag{3-4}$$

Here, G is the cost function, M indicates the muscle forces, $f_i^{(M)}$ is the ith muscle force, $n^{(M)}$ is the number of muscles, A_i is the physiological cross-sectional area of the ith muscle, and N_i denotes the strength of the muscle. C is the coefficient matrix of equilibrium equations, f is the vector of muscle and joint forces, and r is the vector of the external forces and inertial forces.

The inverse dynamics analysis is divided into two steps: a) kinematic fit; b) muscle force calculation, as shown in Fig. 3-4. The muscle force calculation was conducted after the convergence of the fitting. The driving force required to produce this state was calculated inversely to obtain the muscle force of each lower limb muscle in the gait through the fitted gait and the data input in the dynamometer by inverse dynamics analysis.

3.3 Gait coordination analysis methods

Gait coordination is an important indicator of the stability and economy of gait

(Bruijn et al., 2011). In this study, the quantitative analysis of the gait coordination will be performed through 1) inter-joint coordination; 2) decomposition index; 3) muscle coordination. For the definition of the gait cycle and the direction of motion, please refer to Section 2.1 in Chapter II.

3.3.1 Inter-joint coordination

This study investigated gait coordination through the coordination of lower limb joints. Inter-joint coordination study is based on phase portrait of joint motion with the joint angle as X-axis and the joint angular velocity as Y-axis, the angle and velocity of the joint at each moment in the gait cycle expressed in the coordinates, as shown in Fig. 3-6.

Fuchs et al. believe that it is necessary to normalize the phase portrait so that the graph becomes an annulus surrounding the phase plane (Fuchs et al., 1996). In this study, normalization was performed as follows: the joint angle was normalized by the equation (3-5). Therefore, the maximum flexion angle and the extension angle which can be converted from the range of variation of the joint angle are -1 and 1 respectively. The value of 0 represents the middle value of the change of joint angle. The advantage of using this normalization method for joint angles is that the differences in the definition of joint angle (such as the definition of positive and negative) in different studies can be avoided; at the same time, the difference caused by the amplitude and frequency of motion can be reduced. In addition, the attachment of the mark point may cause the difference of the neutral position, the errors can be avoided in the normalization process. The joint angular velocity was normalized using equation (3-6). The nomalized method for angular velocity limits the data to either -1 or 1, depending on the maximum value of the absolute value

of $\dot{y}(t_i)$. The advantage of this normalization process is to preserve the physical meaning of the 0 value so that the feature points with an angular velocity of 0 can be traced back.

$$g(y(t_i)) = 2(\frac{y(t_i) - \min(y(t))}{\max(y(t)) - \min(y(t))}) - 1 \qquad (3\text{-}5)$$

$$f(\dot{y}(t_i)) = \frac{\dot{y}(t_i)}{\max(|\dot{y}(t_i)|)} \qquad (3\text{-}6)$$

Among them, (t_i) represents each moment in the gait cycle, $y(t_i)$ is the lower limb joint angle (hip joint, knee joint, and ankle joint) at each moment in the gait cycle, $\dot{y}(t_i)$ is the angular velocity of the lower limb joint at each moment in the gait cycle. $g(t_i)$ and $f(t_i)$ represent new values of lower limb joint angle and joint angular velocity after normalization.
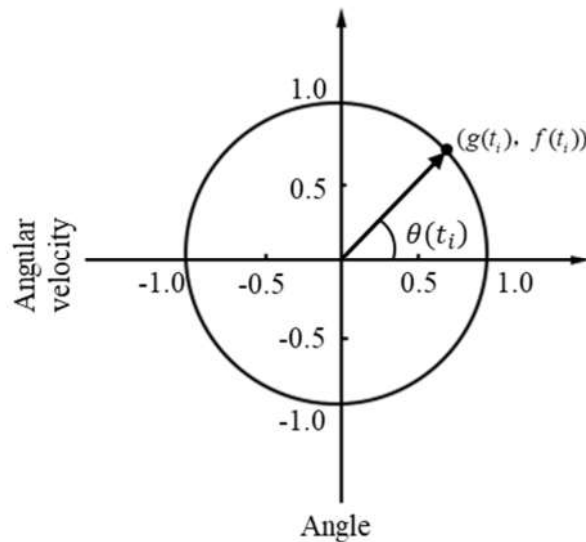
Fig. 3-6 Definition of phase angle θ in phase plane

After the normalization process, a phase angle θ of the joint at each moment is jointly determined by the joint angle and the joint angular velocity by an equation

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(3-7) (Burgess-Limerick et al., 1993). The phase angle in the phase portrait is shown in Fig. 3-6.

$$\theta(t_i) = \arctan(\frac{f(t_i)}{g(t_i)})$$ (3-7)

Inter-joint coordination is an indicator of gait coordination, which is quantified using Continuous Relative Phase (CRP) (Lamb and Stöckl, 2014). CRP is the difference between the phase angle of the proximal joint and the distal joint at the during the gait, as shown in the equations (3-8). A CRP of 0° indicates that the two joint movements are completely synchronized; a positive value indicates that the proximal joint motion is ahead of the distal joint motion and vice versa; a CRP of ±180° means that the movements of two joints are anti-phase. The CRP of hip-knee, knee-ankle and hip-ankle joint were studied. The amputated limbs, non-amputated limbs and the lower limbs of the control group were conducted in this study.

$$CRP(t_i) = \theta_1(t_i) - \theta_2(t_i)$$ (3-8)

$\theta_1(t_i)$ and $\theta_2(t_i)$ represent the phase angles of the two joints in the comparison in the gait cycle respectively.

Inter-joint coordination differences were compared by the following parameters: Cross-correlation (CC), as shown in equation (3-9); Root Mean Square (RMS), as shown in equation (3-10); and Deviation phase (DP), as in equation (3-11) and (3-12).

$$CC = \frac{\sum_{i=1}^{n}(CRP_1(t_i) - \overline{CRP_1(t)})(CRP_2(t_i) - \overline{CRP_2(t)})}{\sqrt{\sum_{i=1}^{n}(CRP_1(t_i) - \overline{CRP_1(t)})^2(CRP_2(t_i) - \overline{CRP_2(t)})^2}}$$ (3-9)

$$RMS = \sqrt{\frac{\sum_{i=1}^{n} CRP(t_i)^2}{n}} \qquad (3\text{-}10)$$

$CRP_1(t_i)$ and $CRP_2(t_i)$ are the CRP data of the two groups; $\overline{CRP_1(t)}$ and $\overline{CRP_2(t)}$ represent the mean values of the data of two groups respectively; n represents the amount of the discrete data in a gait cycle.

The similarity of CRP is described jointly by CC value and RMS difference. The CC value is compared by the CRP data of two groups in the comparison according to equation (3-9), reflecting the correlation between the CRP data of the two groups. The closer the CC value is approaching 1, the more positive correlation the CRP data of the two groups is; the closer to -1, the more negative correlation the CRP data is presented; the value of 0 represents the correlation is 0. RMS is obtained through the calculation of each discrete data point in a set of CRP mean data, reflecting the size and the degree of change of the relative phase of the two joints in the group, which is used to analyze the phase relationship of the two joints, that is, the dephasing (Chiu et al., 2010) .The smaller RMS difference between the CRP data of two groups in the comparison is, the more similar the coordination relationship of the CRP will be. In summary, the closer the CC values of the data of the two groups in the comparison approach 1 and the closer the RMS difference approaches 0, the more similar the CRP will be. In the study, a hypothetical premise is that we regard the gait of normal people as the healthy gait. The smaller the CC value of the amputee's joint coordination is when compared with that of the normal person and the bigger the RMS difference is, the smaller the similarity will be, which indicates that the amputee's gait coordination disorder is more serious (Stergiou et al., 2001).

The variability of inter-joint coordination is evaluated by the average of all discrete point standard deviations of CRP data in each amputee's gait cycle, namely the DP value, as in equation (3-12). DP value represents the degree of variability of inter-joint coordination of the gait among different subjects. A low DP value means less variability in amputee's coordination; a larger DP value means a greater difference in coordination among amputees, reflecting a greater difference among amputees in the control of the center of body weight in the sagittal plane.

$$SD_j = \sqrt{\frac{\sum_{i=1}^{n}(CRP(t_i) - \overline{CRP(t)})^2}{(n-1)}} \qquad (3\text{-}11)$$

$$DP = \frac{\sum_{j=1}^{m} SD_j}{m} \qquad (3\text{-}12)$$

SD_j is the standard deviation obtained from the CRP discrete data of each amputee's gait; m is the number of subjects.

Coordination variability is an intrinsic feature of functional motion control that is a manifestation of the ability to change or adjust motion in order to meet movement requirement (Chiu et al., 2015). In order to adapt to the motion control characteristics of different amputees, a certain range of variability is necessary (Scholz, 1990). Therefore, there is a certain difference in the gait among different normal people, that is, the DP value is within a certain range. Compared with the DP difference among normal people, excessively low DP value is considered to be insufficient flexibility, such as the knee coordination of the elderly (Kemoun et al., 2002); excessively high DP value means instability of the joint control strategy, which is a manifestation of pathological motion control such as injuries, diseases,

etc. (Hamill et al., 1999).

3.3.2 Decomposition Index

Inter-joint coordination analysis will also be discussed accompanied with decomposition movement. Decomposition movement refers to the movement of one joint is pauses while the other joint continues moving (Bastian et al., 1996). The decomposition movement is represented by the decomposition index, which is defined as the proportion of time of decomposition movement in the gait cycle (Earhart and Bastian, 2001). When the angular velocity of a joint is lower than 5°/s, the movement is regarded to paused, that is, the joint is fixed. In this study, the decomposition index of three groups of joints of the lower extremities: hip-knee, knee-ankle, and hip-ankle joints were analyzed. The decomposition movement is considered to be a manifestation of the compensatory mechanism instead of the physical movement defect. If the movement of a joint is affected, decomposition movement will be generated with the other joint as compensation in order to achieve the movement task and maintain accuracy. This compensation mechanism will reduce the joint moment of the movement (Bastian et al., 1996) and make the static joint become "rigid", so the degree of freedom of control is reduced and movement control becomes simpler and easier. Therefore, the decomposition movement is also considered to be a control strategy to reduce the degree of freedom in complex movements of multiple joints. In this study, the increase in the decomposition movement of amputated amputees reflects the change of amputee's motion control regulation, which is the embodiment of the reduction of exercise ability and the regulation of gait coordination by compensatory movement.

3.3.3 Muscle coordination

Muscle coordination is another demonstration of gait coordination. In this study, the calculation of muscle force of non-amputated limbs relied only on the experimental data on the same side, it was only based on the musculoskeletal model and the kinetics collected from force platforms. The muscle recruitment in inverse dynamics is the process of determining the set of muscle forces that will balance the ground reaction force in the rigid-body dynamical system. The calculation is not relevant to the internal forces such as the interaction between different body segments. Although the muscle of the residual limb and the mechanical structure of the prosthesis of the other side are very different from the musculoskeletal system of normal people, they are not involved in the calculation.

This study focuses only on the movements in the sagittal plane, so the study of lower limb muscles also focuses on the flexors and extensors of the hips, knees, and ankles. In addition, this study only focuses on the muscle force of the non-amputated limbs, the muscle force of the residue limbs was not analyzed. According to the definition of Perry (Perry et al., 2010), a total of 24 flexors or extensors of the lower extremities were extracted in this study, as shown in Table 3-3. The muscle bundle represents the number of muscle bundles of each muscle in the musculoskeletal model (Chiu and Chou, 2012b), the discussion of coordination of the stance phase and the swing phase will be discussed separately

Table 3-3 Extracted lower limb muscles by inverse dynamics analysis (Vicon®, 2010)

Muscle Classification	Name of Muscle	The Amount of Muscle Bundles	Function
Hip Flexor	Adductor Longus	6	Extorting and adducting thigh
	Rectus Femoris	2	Contracting to straighten the knee joint and flexing the thigh
	Gracilis	2	Adducting and intorting hip joint
	Sartorius	2	Flexing hip, knee, extorting and abducing thigh, intorting calf
	Iliacus	9	When the proximal support is used, its pulling force is from the bottom to the upfront and flexing the thigh when contracting.
Hip extensor	Biceps Femoris Caput Longum	1	Stretching the thigh, flexing the calf, and rotating the calf externally
	Semitendinosus	1	Stretching the hip joint, flexing the calf when contracting and rotating the calf medially
	Semimembranosus	1	Stretching the hip joint, flexing the knee joint and rotating it medially
	Adductor Magnus	13	Adducting and extorting the thigh when contracting
	Gluteus Maximus	12	Stretching and extorting the thigh

Muscle Classification	Name of Muscle	The Amount of Muscle Bundles	Function
Knee flexor	Biceps Femoris Caput Breve	3	Stretching the thigh, flexing the calf, and rotating the calf laterally
	Popliteus	2	Flexing the knee joint rotating the calf medially
	Gastrocnemius	2	Plantar flexion of ankle and knee joint
Knee extensor	Vastus Lateralis	8	Pulling the patella outward when contracting, straightening the knee joint
	Vastus Medialis	10	Participating in the extension of the knee joint, pulling patella inward when contracting, fixing patella and offsetting the tension outside the patella
	Vastus Intermedius	6	Pulling the patella upward
	Rectus Femoris	2	Straightening the knee joint and flexing the thigh when contracting
Ankle plantar flexor	Soleus	6	Rotating the instep to raising the foot
	Gastrocnemius	2	Plantar flexion of ankle joint and knee flexion
	Tibialis Posterior	6	Plantar flexion, extorsion and adduction of the feet
	Flexor Digitorum Longus	3	Plantar flexion of ankle joint, flexing 2-5 toe, assisting stephenopodia

Muscle Classification	Name of Muscle	The Amount of Muscle Bundles	Function
	Flexor Hallucis Longus	3	Plantar flexion of ankle joint, flexing hallex, assisting stephenopodia
	Peroneus	9	Plantar flexion of ankle joint
Ankle back flexor	Tibialis Anterior	3	Stretching the ankle joint and making stephenopodia
	Extensor Hallucis	3	Stretching the hallux and dorsiflexing the foot and varus
	Extensor Digitorum	3	Extending 2-5 toe, dorsiflexing ankle joint

CHAPTER IV GENERAL CHARACTERISTICS OF GAIT COORDINATION WITH TRANSFEMORAL AMPUTATION

This chapter presents the gait coordination study carried out with transfemoral amputees. The gait analysis was based on wearing the traditional passive transfemoral prosthesis (refer to Section 1.3 of Chapter I), which is the most widely used type of transfemoral prosthesis. The gait coordination was compared with that of abled persons so as to have a better knowledge of the motion control strategies of the amputees.

All eight recruited amputees participated in the research described in this chapter. In the course of the experiment, experiment A (refer to Section 3.1.3) was carried out, in which the amputees walked at the preferred speed with which they feel comfortable. Eight healthy adults served as a control group to conduct the same experiment for reference and comparison.

The study in this chapter was focused on the exploration of the common characteristics of amputees' gait coordination. For this reason, the amputees who participated in this experiment all used the prostheses that they wore in daily life and had been completely adapted. The information of the amputees and the prostheses that they used is presented in Table 3-2. The involved prosthetic knees are products of three brands with different models, and two types of knee joint damping mechanism were used: hydraulic and pneumatic. The common feature of such prostheses is that the knee joint cannot generate an extension moment actively. Thus, when supporting the body weight, the knee joint will stretch and lock up; then,

the joint cannot rotate so as to ensure stability and safety of the body. The body center of gravity is supported to move forward by the lower limbs like an inverted pendulum. Such prostheses rely on the residual limb and the hip joint to drive the prosthesis forward during the swing phase. The prosthetic feet involved in this experiment include the SACH, multi-axis ankle joint, and ESR feet. The working mechanisms of these types of prosthetic feet vary greatly from each other, and the greatest difference lies in the rotation characteristics of the ankle joint or foot elasticity in the stance phase. Among them, the ankle joint of SACH is a fixed joint, there is merely partial elasticity in the heel and forefoot of the whole prosthetic foot. In the multi-axis prosthetic foot, the ankle joint can rotate on multiple axes, and there may be dorsiflexion of the ankle joint in the stance phase. The ESR foot can store the energy by compression into deformation at the stance phase and release it at the end of the stance phase to make up for part of the propulsive forces. In spite of the numerous differences in the products, the common feature of such prosthetic feet is that the ankle joint cannot rotate actively so that there is no dorsiflexion or plantarflexion in the swing phase (refer to Section 1.3.3 in Chapter I).

A statistical analysis was conducted to compare the gait coordination of amputees to abled persons. The spatio-temporal parameters (walking speed, time ratio of the stance phase), kinematic parameters (joint angle, joint angular velocity), dynamic parameters (joint moment), gait coordination analysis parameters (CRP, DP, and DI) were involved (for the parameters, refer to Section 3.3 in Chapter III). The difference between the lower limbs of the amputees (both the amputated and non-amputated limbs) and those of the control group was tested by one-way ANOVA and the parameters were tested by the LSD post hoc (Bastian et al., 1996; Chiu et al., 2013; Chiu et al., 2015; Chiu and Chou, 2012b; Chiu et al., 2010; Earhart and

Bastian, 2001; Hutin et al., 2012; Morton and Bastian, 2003; Yen et al., 2009). The significance level was set at $\alpha = 0.05$.

4.1 Results of gait analysis
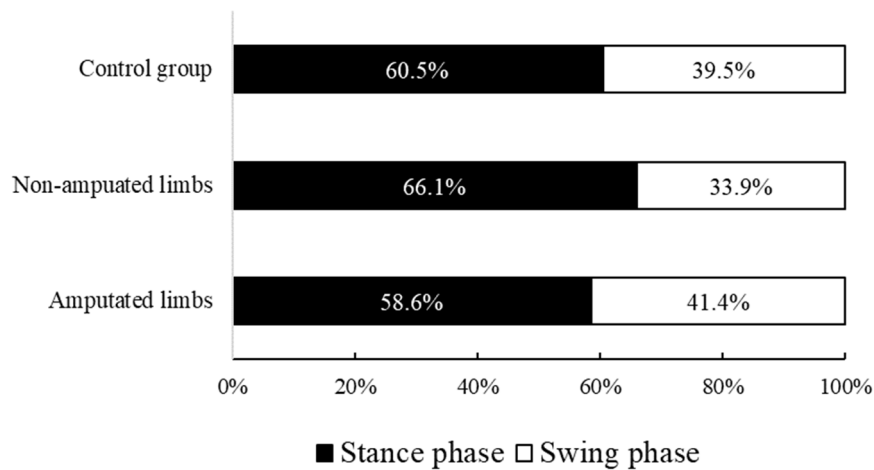
Fig. 4-1 Duration ratio of stance and swing phases for the amputees' amputated limbs, non-amputated limbs, and that of control group in gait. The horizontal ordinate represents the gait cycle

Compared with the walking speed of the control group (1.37 ± 0.13 m/s), the walking speed of the amputated limbs (0.92 ± 0.07) and that of the non-amputated limbs (0.91 ± 0.08 m/s) of the amputees were significantly lower, and the p values were both less than 0.01. The ratio of the stance phase in the control group was 60.5% ± 1.7%, which was close to the golden ratio proposed by Iosa (Iosa et al., 2013b), as shown in Fig. 4-1. However, the stance phase of the amputee's non-amputated limb was significantly longer (p < 0.01), accounting for 66.1% ± 2.6%. On the contrary, the stance phase of the amputated limb was significantly shorter (p < 0.01), accounting for 58.6% ± 1.3%.

The angles of the hip, knee, and ankle joints of the lower limbs of amputees and the

control group are shown in Fig. 4-2 a, b, c. For the definitions of joint angle and joint moment, refer to those in the book "*Whittle's Gait Analysis*" (Whittle, 2003). There was no significant difference between the two lower limbs of amputees and the control group in terms of the hip maximum extension angle (occurring at the end of stance phase) and the hip maximum flexion angle (occurring at the initial contact or the end of the swing phase) ($p > 0.05$). The joint angle of the prosthetic knee in the stance phase changed slightly, and the maximum flexion angle before the pre-swing phase (the flexion extreme point in Fig. 4-2 b was approximately 4.4° ± 4.7°) was significantly less than that of the control group (12.6° ± 6.2°, $p < 0.01$) and that of the non-amputated limbs (11.4° ± 8.2°, $p < 0.01$). There was no significant difference in the maximal flexion angle (occurring at the middle of the swing phase) of the knee joints of the three groups in the gait cycle, all of which were approximately 59.9° ± 9.1°. The maximum dorsiflexion angle of the ankle joint occurred at the end of the stance phase. There was no significant difference between the maximum dorsiflexion angle of the amputees' prosthetic ankles and that of the control group, but the maximum dorsiflexion angle of the amputees' non-amputated limbs (6.8° ± 5.0°) was significantly less than that of the control group (11.2° ± 3.2°, $p = 0.05$). In addition, because of the passivity of the prosthetic joint, the prosthetic ankle rotated by almost 0° at the swing phase, while there was no significant difference between the amputees' non-amputated limbs and the limbs of the control group.
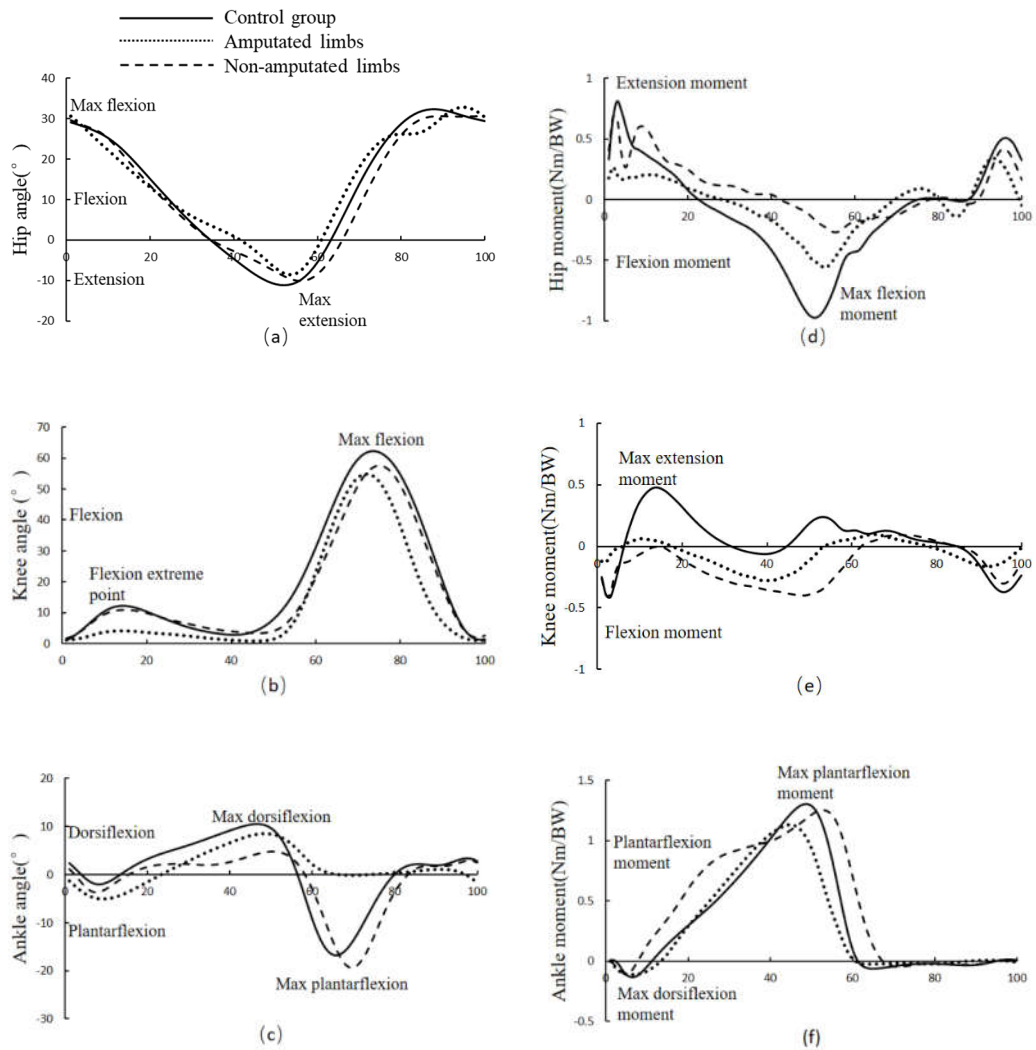
Fig. 4-2 Comparison of the ensemble mean curves of: (a) hip angle; (b) knee angle;

(c) ankle angle; (d) hip moment; (e) knee moment; (f) ankle moment within the 100%

gait cycle. BW: body weight

The moments of the hip, knee, and ankle joints of the amputees and the control

group in the gait cycle are shown as Fig. 4-2 d, e, f. The maximal flexion moment

of the hip joints in the control group (-1.0 ± 0.5 Nm/BW) were significantly higher

than those of the amputated limbs (0.7 ± 0.2 Nm/BW, $p = 0.46$) and non-amputated

limbs (-0.4 ± 0.3 Nm/BW, $p < 0.01$) of the amputees. As for the knee joint moment,

the control group mainly provided extension moments in the stance phase, while

the knee joints of the amputated and non-amputated limbs hardly produced any

extension moments, and mainly produced flexion moments. During the swing phase, the knee joint moments of the three groups showed similar trends, but there were significant differences at the end of the swing phase. The greatest difference occurred at around 96% gait cycle, where the value of the control group (-0.4 ± 0.1 Nm/BW) was significantly higher than that of the amputated limbs (-0.2 ± 0.01 Nm/BW, $p < 0.01$) and the non-amputated limbs of the amputees (-0.3 ± 0.1 Nm/BW, $p = 0.42$). There was no significant difference in the maximum dorsiflexion moment and the maximum plantarflexion moment of the ankle joint between the three groups. However, it is noteworthy that the duration of the plantarflexion moment produced by the ankle joints of the non-amputated limbs accounted for $56.9\% \pm 4.7\%$ of the whole gait cycle, which was significantly higher than that of the amputated limb ($43.8\% + 7.9\%$, $p < 0.01$) and that of the control group ($46.3 \pm 3.6\%$, $p < 0.01$), while there was no significant difference between the amputees with amputated limbs and the control group.

(a)

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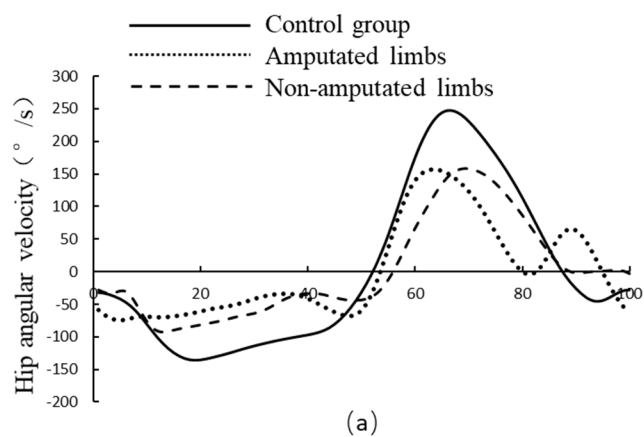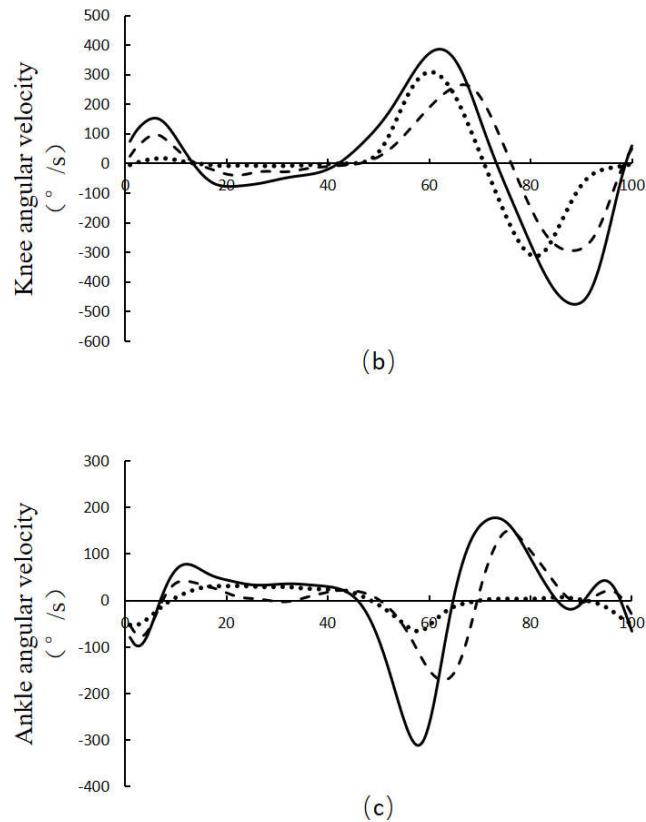
Fig. 4-3 Comparison of the joint angular velocity within the 100% gait cycle: (a) hip joint; (b) knee joint; (c) ankle joint

Although the hip and ankle joints of the amputated limb were not significantly different from those of the control group in joint angle and joint moment at the stance phase, it was found that the hip angular velocity of amputated limbs was significantly lower than that of the control group during the period from about 23% gait cycle to 39% gait cycle in the stance phase ($p < 0.05$), with an average decrease of approximately −69.7°/s. The amputees with prosthetic ankles showed significant differences from the control group in the period from 9% gait cycle to 17% gait cycle with an average decline of approximately 46.7°/s, and in the period from 52% gait cycle to 63% gait cycle with an average decline of −199.8°/s in the pre-swing phase ($p < 0.05$). There was no significant difference in the joint angle and joint moment between the knee and ankle joints of the non-amputated limbs, but from

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the analysis of joint angular velocity, it was found that the knee angular velocity of amputees with non-amputated limbs was significantly lower than that of the control group in the middle swing phase from around 72% gait cycle to 80% gait cycle, with an average decrease of approximately −115.7°/s. The angular velocity of the ankle joints of the amputees with non-amputated limbs was significantly different from that of the control group at the beginning of the swing phase from 65% gait cycle to 71% gait cycle, with an average decrease of approximately 165.3°/s.

The hip angular velocity of the amputees' non-amputated limbs was 1.1 ± 2°/s on average in the last 5% gait cycle of the swing phase, which was significantly less than −32.9 ± 5.4°/s in the control group. In addition, non-amputated limbs' knee angular velocity in middle swing phase, amputated limbs' hip angular velocity in middle swing phase and terminal swing phase, and amputated limbs' knee angular velocity in middle swing phase were all found discrepancies compared to control group.

The inter-joint coordination, which is the quantization parameter of gait coordination, of the hip–knee, knee–ankle, and hip–ankle groups was calculated as CRP in the gait cycle, as shown in Fig. 4-4. When the CRP value is positive, it means that the motion of the proximal joint is ahead of that of the distal joint, and vice versa. A CRP of 0° means that the motions of the two joints are synchronized completely, whereas when the CRP is ±180°, it means that the motions of the two joints are in inverse phase.
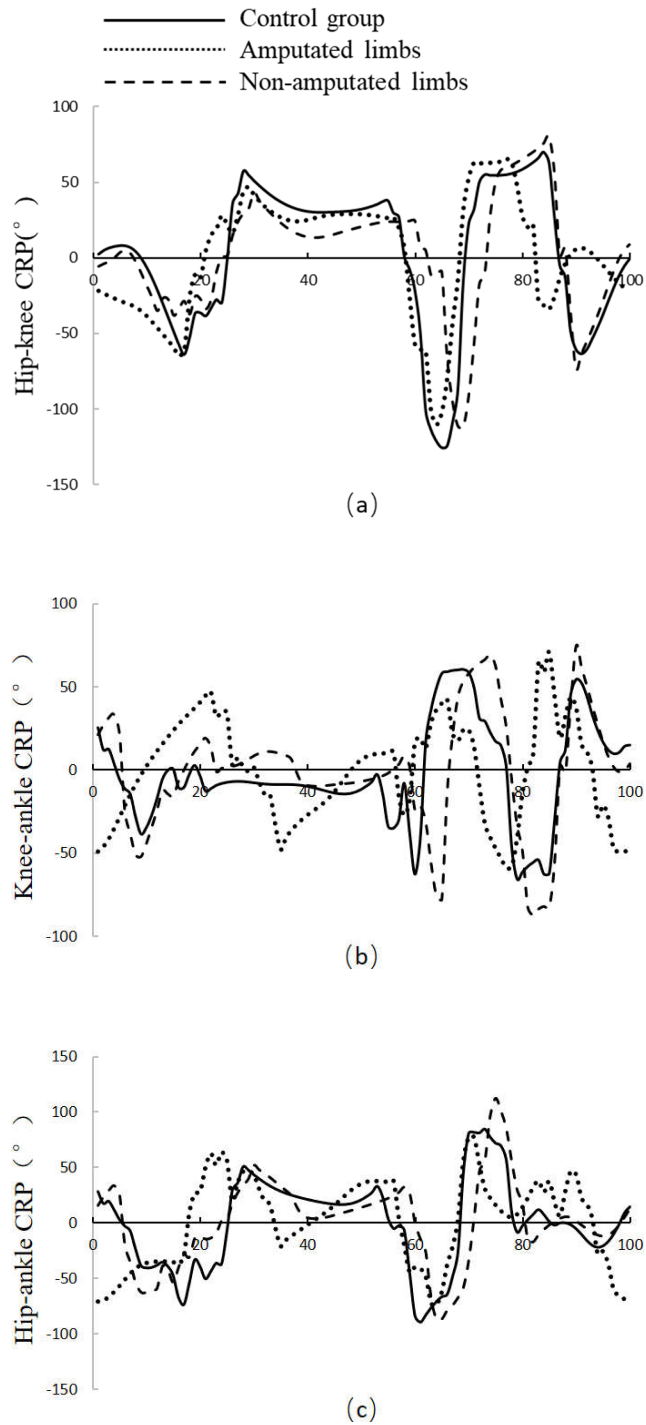
Fig. 4-4 Comparison of ensemble mean CRP within the 100% gait cycle: (a) hip–knee; (b) knee–ankle; (c) hip–ankle

The mean CRP of amputated and non-amputated limbs of the amputees was correlated with that of the control group, respectively, as presented in Table 4-1.

Whether in the stance or swing phase, the amputees' hip–knee coordination of the non-amputated limbs demonstrated a high correlation with that of the control group, and the CC value was 0.89 in the stance phase and 0.93 in the swing phase, respectively. While the hip–knee coordination of the amputated limbs was less correlated with that of the control group in both the stance and swing phases, the CC values were both 0.79. There was a great difference in the knee–ankle CRP correlation of the two lower limbs of the amputees when correlated with the control group. The amputated limb had a very low correlation with the limbs of the control group, amounting to −0.05 in the stance phase and 0.23 in the swing phase, respectively. The correlation between the non-amputated limb and the limbs of the control group was smaller in the stance phase, 0.78, and higher in the swing phase, 0.89. The hip–ankle CRP of amputated limbs was less correlated with that of the control group, with a CC value of 0.30 in the stance phase and 0.66 in the swing phase. However, the hip–ankle CRP of the non-amputated limb still showed a high correlation with that of the control group, with a CC value greater than 0.8 in both the stance and swing phases.

Table 4-1 Cross-correlation coincidence (CC) for the ensemble mean CRP of the amputated limbs, non-amputated limbs of amputees to the control group

Joint groups	Gait phase	Amputated limbs-Control group	Non-amputated limbs-Control group
Hip-knee	Stance	0.79	0.89
	Swing	0.79	0.93
Knee-ankle	Stance	−0.05	0.78
	Swing	0.23	0.89
Hip-ankle	Stance	0.30	0.81
	Swing	0.66	0.88

Table 4-2 RMS of the ensemble mean CRP

Joint groups	Gait phase	Control group	Amputated limbs	Non-amputated limbs
Hip-knee	Stancee	44.8 ± 4.4	43.2 ± 2.6	46.2 ± 7.3
	Swing	70.4 ± 6.8	66.6 ± 9.7	69.2 ± 8.7
Knee-ankle	Stance	33.4 ± 5.8	40.5 ± 6.2	36.6 ± 12.8
	Swing	56.1 ± 7.7	58.9 ± 18.6	66.6 ± 5.9
Hip-ankle	Stance	46.2 ± 10.1	55.8 ± 7.0	52.5 ± 5.8
	Swing	53.8 ± 14.9	65.1 ± 11.2	61.2 ± 11.4

The RMS of the mean CRP values in the amputees' amputated limbs and non-amputated limbs and the limbs of the control group are listed in Table 4-2. The difference in the RMS of the hip–knee CRP of both the amputated （66.6 ± 9.7) and non-amputated limbs (69.2 ± 8.7) was smaller than that of the control group in the stance phase, and they were 3.5% lower and 3.2% higher than that of the control group, respectively. The amputated limb had a 5.4% difference (66.6 ± 9.7 vs. 70.4 ± 6.8) in the RMS of the hip–knee CRP from that of the control group in the swing phase, but the difference between the non-amputated limbs and the limbs of the control group was still very low at the swing phase, just amounting to only 1.7% (69.2 ± 8.7 vs. 70.4 ± 6.8). The difference in the RMS of the knee–ankle CRP between the amputated limbs and the limbs of the control group was as high as 21.2% (40.5 ± 6.2 vs. 33.4 ± 5.8) in the stance phase, but it decreased to 5% in the swing phase (58.9 ± 18.6 vs. 56.1 ± 7.7). The knee–ankle CRP of the non-amputated limb was also significantly different from that of the control group. However, unlike the amputated limb, the difference in knee–ankle CRP between the non-amputated limb and the limbs of the control group reached the highest at the swing phase, amounting to 18.6% （66.6 ± 5.9 vs. 56.1 ± 7.7）, whereas the difference in the

stance phase was smaller, amounting to only 9.3% (36.6 ± 12.8 vs. 33.4 ± 5.8). The differences between the hip–ankle CRP of the amputated limb and that of the control group were both higher than 20% in the stance and swing phases. Although the difference in the hip–ankle CRP between the non-amputated limb and the limbs of the control group was smaller, there was still a difference of more than 13% at both the stance and swing phases.

The variation in CRP is represented by DP, as shown in Fig. 4-5. The variations in hip–knee, knee–ankle, and hip–ankle CRP of the non-amputated limbs had no significant differences from those of the control group, whether in the stance or swing phase ($p > 0.05$). The amputated limb had a different performance: the hip–knee DP was similar to those of the control group in both the stance and swing phases ($p > 0.05$); the knee–ankle DP was significantly higher than those of the control group (40.7 ± 6.3 vs. 31.6 ± 4.7, $p = 0.046$) in the stance phase only; the hip–ankle DP was also significantly higher than that of the control group in the stance phase (52.6 ± 7.0 vs. 46.4 ± 10.2, $p = 0.02$), while there was no significant difference in the swing phase ($p > 0.05$).
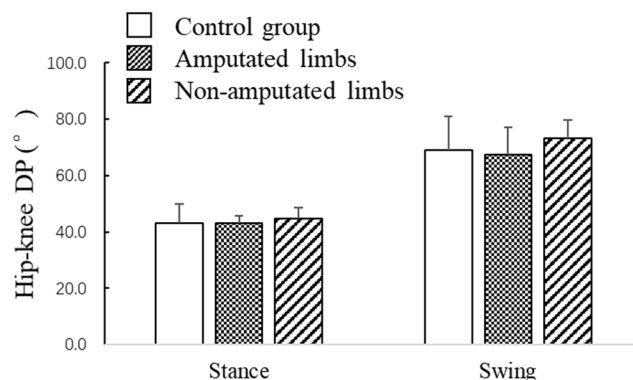
(a)
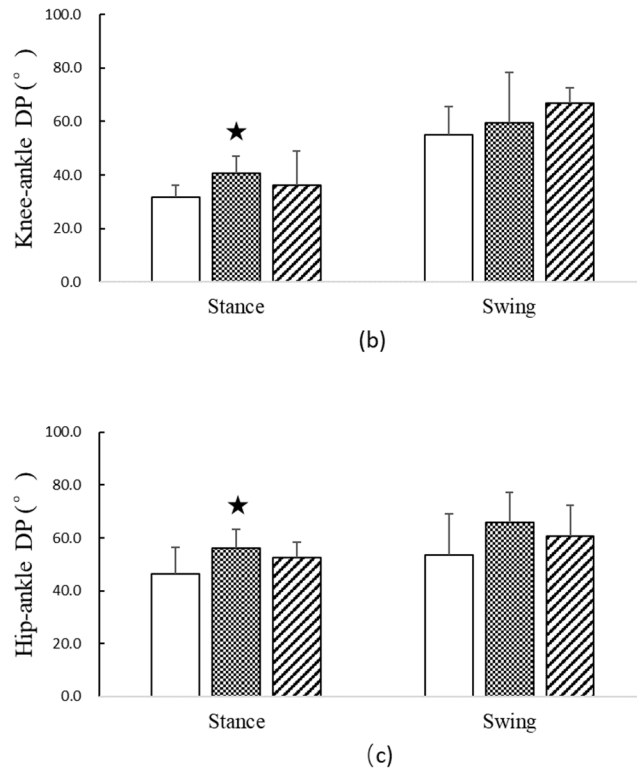
Fig. 4-5 Comparison of the DP among the amputated and non-amputated limbs of amputees and the limbs of the control group: (a) hip–knee; (b) knee–ankle; (c) hip–ankle; ★ indicates that there is a significant difference in DP value between amputees and the control group.

Through a statistical analysis of the DI of amputees and the control group, it was found that the DI of amputees was very different from that of the control group, especially in the amputated limbs, as shown in Fig. 4-6. The hip–knee DI of the amputated limb was approximately 42.4 higher than that of the control group in the stance phase (63.3 ± 20.9 vs. 20.9 ± 19.5, $p < 0.01$), and 11.8 higher in the swing phase (20.1 ± 7.5 vs. 8.3 ± 4.7, $p = 0.02$). There was no significant difference between the non-amputated limb and the limbs of the control group in the stance phase. However, in the swing phase, it was 17.2 higher than that of the control group (25.5 ± 16.3 vs. 8.3 ± 4.7, $p = 0.03$). The knee–ankle DI of the amputated limbs of

amputees was significantly different from that of the control group and the non-amputated limbs of amputees in both the stance and swing phases. It was 45.2 higher than that in the control group in the stance phase (63.9 ± 10.7 vs. 18.7 ± 12.9, $p < 0.01$), and 50.6 higher than that in the swing phase (64.3 ± 20.1 vs. 13.7 ± 8.1, $p < 0.01$). The knee–ankle DI of the non-amputated limbs was similar to that of the control group in both the stance and swing phases. The hip–ankle DI of the amputated limbs was not significantly different from that of control group in the stance phase, but it was 44.1 higher than that of the control group in the swing phase (59.7 ± 20.3 vs. 15.6 ± 10.9, $p < 0.01$). The hip–ankle DI of the non-amputated limbs of amputees was 9.6 higher than that of the control group in the stance phase (22 ± 11.2 vs. 12.4 ± 13.2, $p=0.04$), while there was no significant difference in the swing phase.

(a)

(b)
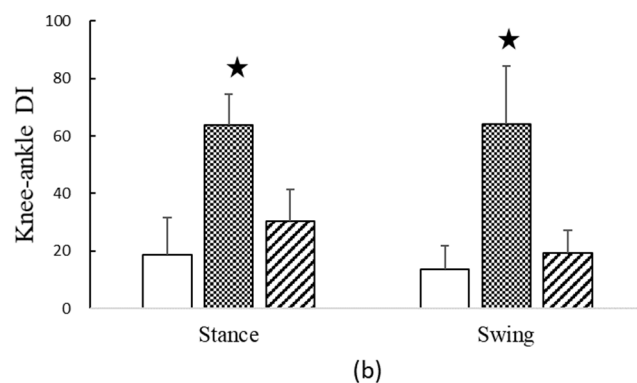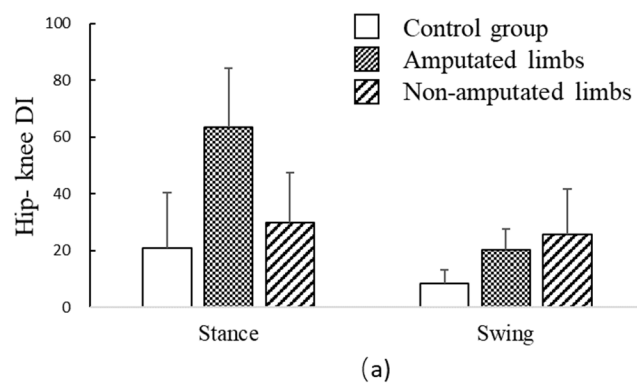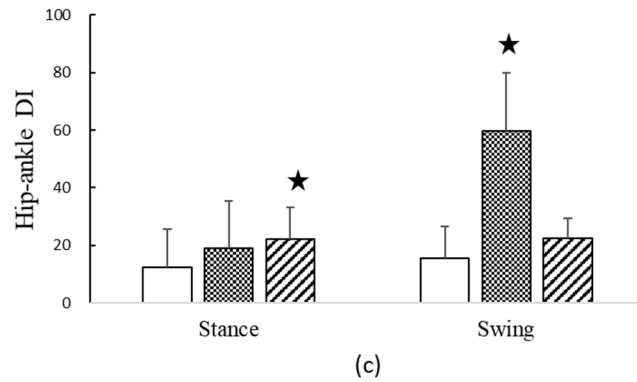
Fig. 4-6 Comparison of the decomposition index (DI) among the amputated and non-amputated limbs of amputees and the limbs of the control group: (a) hip–knee; (b) knee–ankle; (c) hip–ankle; ★ indicates that there is a significant difference in DI between amputees and the control group.

This chapter presents the study of the general characteristics of gait coordination of unilateral transfemoral amputees. The amputees wore the traditional passive prostheses. Gait coordination of the lower limbs was analyzed and compared with that of abled persons to identify the differences. The gait coordination was represented by inter-joint coordination index, which was quantified by the CRP. The DI was also considered for the discussion of the compensatory mechanism of amputees.

4.2 Conventional gait analysis

Previous studies have found differences between the gait of unilateral amputees and that of abled persons (Sapin et al., 2006, 2008; SilverThorn et al., 2009; Torrealba et al., 2008; Versluys et al., 2009). This work also compared the parameters studied in previous studies, including walking speed, time ratio of the stance phase, joint angle, and joint moment, and obtained identical results. For example, the gait of the

amputated and non-amputated limbs of the amputee was different from that of the control group in the spatio-temporal, kinematic, and dynamics parameters (Fig. 4-2). To ensure comfort and stability in walking, the walking speed of the amputees was approximately 0.2 m/s slower than that of the control group. Compared with that of the control group, the time ratio of the stance phase of the amputated limb of the amputee was approximately 1.9% smaller, while that of the non-amputated limb was about 5.6% higher. The hip joint angles of the amputated and non-amputated limbs of the amputees showed similar performance to those of the lower limbs of the control group, but the maximum hip flexion moments of amputated and non-amputated limbs showed a decline of 0.3 Nm/BW and 0.6 Nm/BW, respectively. The non-amputated limb knee joint showed a similar joint angle pattern to that of the control group, but its joint moments demonstrated a larger flexion moment or a smaller extension moment in the whole stance phase. Although the maximum ankle dorsiflexion and plantarflexion angles of the non-amputated limb were similar to those of the control group, the duration of ankle plantarflexion moment was approximately 10.6% gait cycle longer than that of the control group. In addition, the performances of the prosthetic knee and ankle joints were significantly different from those of the control group in terms of kinematics and dynamics because of their prosthetic mechanisms.

By comparing the above conventional parameters, we can find many gait differences between amputees and abled persons. The rehabilitation of amputees and the performance verification of prostheses are usually based on such methods, and these methods also serve as an indicator of whether the gait of amputees is reasonable (Blumentritt et al., 1998; Boonstra et al., 1994; Boonstra et al., 1996; Dumas et al., 2009; Highsmith et al., 2010; Sapin et al., 2006; Segal et al., 2006;

SilverThorn et al., 2009). However, the working mechanism of the prosthetic product cannot fully compensate the complex biomechanical functions of the lost joints , amputees will produce a compensatory movement during walking to offset the deficiencies of the current prosthesis (Bastian et al., 1996). Taking the prosthetic knee as an example, flexion or extension is allowed only on the sagittal plane in gait, while rotation is restricted on the coronal and transverse planes to ensure gait stability and safety. It is also owing to the different working modes of the mechanical structures, while it contributes to the gait deviation as well. Therefore, to ensure the stability and accuracy of gait, the control strategies of other parts of the body in gait change accordingly, resulting in a different gait performance. In the analysis of joint parameters, the performance of a single joint can implicate some differences in gait, such as the prosthetic knee angle and the ankle joint angle of the non-amputated limb, and the hip and knee joint moments of the prostheses and the non-amputated limb (Fig. 4-2), but these differences are not sufficient to show the change of the amputees' motion control mode. Amputees may respond to the gait impact of the prostheses' deficiency by changing the motion control strategy of specific joints. Even if the parameters of joint angle and joint moment of some joints are consistent with those of abled persons, the joint performance may be due to the effect of adjustment of other parts of the amputees' body. In this case, the change of coordination between joints, or the compensatory effect of the change of coordination, can more accurately reveal the defects in the gait of amputees.

In clinic, whether the gait of the amputees after rehabilitation training is reasonable and whether the performance of a prosthesis is in good performance should not only depend on the study of individual joints, but also to further analyze the coordination between joints. The optimization of joint control needs to consider the coordination

between the joints, which may help to facilitate the improvement of rehabilitation and prosthesis design

4.3 Similarity between amputees and abled persons in inter-joint coordination

From the perspective of the similarity of inter-joint coordination, the amputated limb' gait was different from that of the abled persons. Amputees tended to change their control strategies over motion so as to ensure the stability and accuracy of gait. By comparing CC and RMS (Table 4-1 and Table 4-2), the hip–knee coordination of the amputated limbs maintained a similarity to that of the control group to a certain extent, in the stance or swing phase, with a CC value above 0.79 and RMS difference less than 6%. The knee–ankle and hip–ankle coordination of amputated limbs was little similar to those of the control group. The hip–knee coordination of amputees' non-amputated limbs showed very high similarity with that of the control group in both the stance and swing phases, with CC values of the stance and swing phases above 0.89 and RMS difference of less than 5%. However, the values of the knee–ankle and hip–ankle coordination of non-amputated limbs were different from those of the control group in different degrees.

The difference in the coordination pattern indicates differences in the control strategies. The passive transfemoral prostheses worn by amputees cannot flex the knee joint as abled persons during the stance phase. Otherwise, amputees are prone to walking unstably and falling. In the same way, the prosthetic ankle cannot rotate actively, and thus, it cannot dorsiflex in the swing phase to increase the height of toes off the ground. The differences of these mechanisms from those of abled

persons affect the hip–knee, knee–ankle, and hip–ankle coordination of the amputated limb. In addition, the prostheses involved in this study were all split type, namely the prosthetic knee and ankle were independent. The prosthetic knee and ankle joints are designed to be fixed or rotated in a relatively fixed way in gait. Thus, the two prosthetic joints work relatively independently from each other in gait, with a lack of cooperation. During walking, the amputated limb of the amputee adjusts the control over motion mostly based on the hip joint and other parts of the body, but it fails to actively adjust its prosthetic knee and ankle joints to cope with different walking conditions. The adjustment of control strategies affects the control of the musculoskeletal system of the lower limbs on the contralateral limb, and that is why the non-amputated limb contains a complete musculoskeletal system but will also change its inter-joint coordination accordingly.

The coordination analysis can also distinguish the differences that are difficult to find in a conventional gait analysis. There was no significant difference in joint angle and joint moment was found between the amputated limbs and those of the control group for the the hip and ankle joints during stance phase (Fig. 4-2). It is generally regarded that the gait of the amputated limb is similar to that of abled persons in such condition. However, in this study, it was found that the hip–ankle coordination of the amputated limb was significantly different from that in the control group in the stance phase, with a CC value of 0.3 and an RMS difference of more than 20%. Similarly, the joint angles and joint moments of the knee and ankle joints of the non-amputated limbs were similar to those of the control group in the swing phase, but there were also significant differences in the knee–ankle coordination with an RMS difference as high as 18.6%.

The coordination analysis in this study expressed the motion of a single joint as a phase angle, which was determined by the joint angle and joint angular velocity at the same time (refer to Fig. 2-6). Although the joint angular velocity is also derived from the first derivative of the joint angle to time, it usually scales the ratio into 100% gait cycle in a conventional gait analysis, which will be conducive to comparing the parameters more conveniently, with certain advantages. At the same time, owing to the scaling of the parameters relative to time, some features are abandoned. For example, angular velocity should be the slope of an angle curve. However, the slope of the normalized curve in conventional gait analysis does not represent the angular velocity. Owing to the differences between the joint angular velocity, the inter-joint coordination analysis distinguished the gait deviation of amputees more sensitively. In addition to the change in motion control of a single joint, inter-joint coordination analysis has the potential to reveal the change in the coordination relationship between the two joints. As a result, gait analysis of single joint can be improved to study the coupling relationship of two joints and quantified to interpret the adjustment of motion control.

4.4 Variations in inter-joint coordination among amputees

The variation represents the differences in gait coordination between amputees wearing different types of prostheses. A low variation indicates that there is a similar gait coordination in the gait of amputees wearing all types of prostheses, whereas an excessive variation means that there are significant differences in the gait coordination of the amputees. The variation is the representation of different control strategies and embodiment of functional differences among the prosthetic limbs.

Except for the differences in the prosthetic knee and ankle joint arising from the

different mechanism, there was no significant difference in the variation of inter-joint coordination in the sound joints of amputees wearing different types of prostheses, compared with that of the control group (Fig. 4-5). Although prosthetic knees and ankle joints can be classified into different catalogues, passive transfemoral prostheses had similar impact on the inter-joint coordination of amputees' gait. There was no significant difference of inter-joint coordination between the non-amputated limbs and the limbs of the control group in the variations (hip–knee, knee–ankle, and hip–ankle), whether in the stance or swing phase. The hip–knee coordination of the amputated limbs was also similar in variation as the control group. The difference between the amputated limbs and the control group in inter-joint coordination were found in knee–ankle and hip–ankle of which the variations of amputees were higher in the stance phase, and there was no significant difference in the swing phase. The prosthetic knee and prosthetic ankle are two passive joints, which fail to be adjusted actively in the face of different walking conditions or even different gait cycles. For this reason, the working mechanism of different types of prosthetic joints has particularly sensitive impacts on the variations of coordination. In this study, the amputees wore several different types of prosthetic feet, such as SACH, multi-axis ankle-foot, and ESR foot (see Table 3-2 in Chapter II). The working mechanisms of these prosthetic feet had great differences in the working mechanism, which is the most important reason for the great variation in the knee–ankle and hip–ankle coordination on the amputated limbs. In spite of the different types of the prosthetic knees, the greatest difference lies in the way of achieving joint damping adjustment (pneumatic or hydraulic), while there are similar working mechanisms but fewer effects on coordination. All the amputees who participated in this study had their limbs amputated for a long

time and were fully adapted to their own prostheses. Thus, even the prostheses were of different models, the difference in inter-joint coordination still reached a normal range as the control group. That is to say, the different type of prostheses used in this study was similar with respect to compensation of the biomechanical function of amputees, and no significant differences were found in gait coordination among amputees. At the same time, the consistency of coordination among amputees also showed that the amputees had been more adapted to their own prostheses after rehabilitation training, which was also in line with our screening criteria for the subjects (See Section 3.1.2 in Chapter III for details).

However, it is worth noting that the consistency of coordination does not mean that the different models of prostheses have exactly the same compensatory effect on the amputees. There may be substantial variations in the impact on the amputee owing to the characteristics of each prosthetic joint. Taking the ESR feet as an example, the design of the elastic foot enables the prosthetic foot to store energy in the first half of the stance phase by bending under pressure, and increases the propelling force from the ground by releasing the stored energy in the later stage of the stance phase, thus reducing the metabolic expenditure (Fey et al., 2012; Versluys et al., 2009). It is a direct improvement for the failure of amputees with prostheses to produce enough propulsive forces in gait. According to the research of variations of inter-joint coordination among amputees, the results showed that ESR prosthetic feet have no negative impact on the coordination of gait while giving play to their unique functions. This ensures that the gait coordination of amputees will not be affected while the products give play to their own features.

From the discussions on coordination patterns in the preceding text, it was shown

that the inter-joint coordination pattern in amputees' gait is different from that of abled persons. In spite of these differences, the coordination variations between different amputees are within a reasonable range. The difference in the coordination pattern of the amputees is due to the change in the control strategy of motion, which tries to improve the accuracy and stability of motion. However, this change in the control strategy is accompanied by an increase in the burden on the rest of the body and the metabolic expenditure (Russell et al., 2017). Therefore, amputees can achieve smooth and accurate motions through the change in control strategy, but also exert negative effects on other parts of the body.

In addition, previous studies have shown that the variations in the coordination of a repetitive gait in a single amputee may be disturbed by other factors, such as pain (Radin et al., 1991) or perception (Roerdink et al., 2007). The Factors may create an input of abnormal information, which is perceived by the joints and result in significant changes in the variations. In future studies, researchers may analyze the variations in inter-joint coordination of a repetitive gait for individual amputees and discover other pathological problems for clinical rehabilitation.

4.5 Gait decomposition movements of amputees

Amputees offset the biomechanical deficiencies through compensatory mechanisms. The insufficient function of the prosthesis can be compensated by adjusting the control of the intact parts of the body.

According to the DI (Fig. 4-6), the working mechanism of prosthetic joints is different from that of abled persons, which directly contribute to the significant change in the DI of the amputated limb compared to abled persons. In spite of this,

the significant increase in such DI is merely an embodiment of the different working modes of the mechanical structures. Except for hip–ankle coordination in the stance phase, other inter-joint coordination in the amputated limbs showed significant differences from those of the control group. According to the design, the passive prosthetic knee needs to keep the ground reaction force anterior to the rotating center of the knee joint in the stance phase, while the prosthetic knee is locked up so as to ensure security. The lower limbs are supported by the prosthetic limb to move the center of body weight forward like an inverted pendulum. If the center of gravity of the amputee's gait is excessively backward, the force line of the ground reaction force will be on the posterior of the prosthetic knee, which will produce the external moment that makes the joint flex. A distinct feature of the passive prosthetic knee is that it cannot actively provide joint extension moments. The reaction force of the ground in the rear may cause the amputee to fall. Such a working mechanism makes the knee joint change less in the angle at the stance phase (Fig. 4-2 b), which leads to an increase in the hip–knee and knee–ankle DI by 42.4 and 45.2, respectively, at the stance phase. Similar to the prosthetic knee joint, the prosthetic ankle joint tends to be stationary in the swing phase owing to its inability to actively produce dorsiflexion. The DI of the knee–ankle and hip–ankle joints show an increase of 50.6 and 44.1, respectively, in the swing phase compared with those of the control group.

The adjustment of the hip joint of the non-amputated limb plays an important role in guaranteeing gait stability. Researchers have found in previous studies that the adjustment of control strategies of the proximal joints of the body plays a crucial role in ensuring the balance and stability of gait (Chiu and Chou, 2012b; Monno et al., 2002; Winter, 1992). In this study, the hip–knee DI of the non-amputated limbs

of the amputees was 17.2 higher than that of the control group in the swing phase (Fig. 4-6 a). The non-amputated limb in the swing phase indicates that the contralateral limb is in a single support phase. Due to the insufficient biomechanical function, the amputated limb fails to provide sufficient adjustment for the distal joint, so that it is weaker for adjusting the stability of gait. The instability of coordination may affect the swing phase of the lower limb on the opposite side, namely, the non-amputated limb. As a result, the non-amputated limb shortens the gait cycle by approximately 5.6% compared with the limbs of the control group in the swing phase (Fig. 4-1), and enters the double support phase earlier. Thus, it increases the double support phase duration so as to ensure the stability of gait, which leads to gait deviation related to hip joint. From the perspective of the angular velocity of the joint, the average angular velocity of the hip joint of the amputee's non-amputated limb was $1.1° \pm 2°/s$ in the last 5% gait cycle of the gait, which indicates that the hip joint has ended the forward swing of the lower limbs earlier, slowed down the movement, and waited for the heel strike of the next gait cycle. For a similar reason, owing to the changes in angular velocity of the hip and knee joints of the amputated limbs, the hip–knee DI of the amputated limbs had a significant increase compared with that of the control group in the swing phase. The increase in decomposition movement affects the coordination between joints. For this reason, it is necessary to carry out a more targeted athletic training for the hip joint of the non-amputated limb in the process of rehabilitation. Thus, the athletic ability of the joint will be enhanced, which will have a positive impact on amputees' gait coordination.

The adjustment of ankle joint of the non-amputated limb in the stance phase lowered the difficulty of motion control. The hip–ankle DI of the amputees' non-

amputated limbs was approximately 9.6 higher in the stance phase than that of the control group. The foot is the interface between the body and the ground in walking, and it is directly affected by the GRF in gait. The motion of the ankle joint not only affects the effect of GRF on the human body, but also has an important influence on the proximal joints and even on the center of gravity of the body. The gait of the non-amputated limb of amputees is different from that of abled persons. For example, the stance phase was longer (Fig. 4-1), the muscles of lower limbs were stimulated for a longer time (Castro et al., 2014; Cerqueira et al., 2013; Wentink et al., 2013), and the maximum dorsiflexion angle of the ankle joint was approximately 4.4° smaller than that of abled persons (Fig. 4-2 c). Such differences are all linked to the fact that the ankle joint of the non-amputated limb tends to be more rigid in the stance phase. The ankle joint of the non-amputated limb increased the duration of the plantarflexion moment by approximately 13.6% gait cycle compared with that of the control group, and it had been proved that plantarflexors exert muscle forces over a longer time (Castro et al., 2014; Cerqueira et al., 2013; Wentink et al., 2013). The maximum stiffness of the ankle joint may not increase, but the stiffness of the ankle joint resists the dorsiflexion for a longer time. The musculoskeletal system has a high degree of freedom, and there may be numerous possibilities for compensatory strategies of gait, including adjustments of muscle forces, ligaments, and even articular surfaces (Lu et al., 1998), all of which require varying degrees of control change. The ankle joint was adjusted to be more rigid and it produces more decomposition movement with the hip joint, which can effectively reduce the freedom of control by the musculoskeletal system of the body (Bastian et al., 1996; Earhart and Bastian, 2001) and make the adjustment of coordination simpler and more effective.

In addition, a more rigid system of motion can help to push the body forward more easily. The motion of the parts of lower limbs with phase difference will consume more power at the process of power transmission (Donelan et al., 2002; Hutin et al., 2012). The control adjustment of ankle of the non-amputated limb not only ensures the stability and accuracy of gait but also contributes to the reduction of metabolic expenditure. How to reduce the metabolic expenditure in gait has always been an important issue in the prosthesis research, and the knowledge about the decomposition movement will help to have a further understanding of the increase in metabolic expenditure in amputees.

In this study, the amputees took the initiative to adjust the coordination of the ankle joint of the non-amputated limbs. In future rehabilitation studies, it is suggested that the ankle joint of the non-amputated limb could be given active intervention to alleviate the athletic burden of the amputees. For example, researchers may consider increasing the stiffness of the ankle joint by an exoskeleton or by wearing ankle boots to achieve the same effect, thereby reducing the muscular strength of the ankle joint and making it easier for amputees to maintain stability in gait.

4.6 Defects in amputees' athletic ability

Previous studies on the kinematics of transfemoral prosthesis and clinical rehabilitation of amputees have tried to make the amputees' gait closer to that of abled persons but have placed the walking speed and other athletic abilities in a secondary position. Bastian et al. studied patients with cerebellar ataxia and drew the conclusion that slowing down the speed of movement and adopting decomposition moments can increase the accuracy of movement, which is a manifestation of a compensatory gait (Bastian et al., 1996). Reducing the speed of

movement can make the gait more stable and enables the amputees to be more similar to abled persons in gait, which is a compromise for the psychological rehabilitation. The gait of slow walking makes the kinematics defects less evident, but the concealed problems in athletic ability have not yet been solved. It will be of great significance for amputees to improve their ability to control the speed of joint movement and reduce the compensatory gait.

The joint angular velocity is an indicator of motion control ability. In this study, the amputees reduced the speed of joint movement and increased the decomposition movements. It is beneficial to reduce the degree of freedom of movement and ensure a stable and coordinated gait. However, the increase in decomposition movements also led to the decrease in walking speed. The preferred walking speed with which the amputees feel comfortable was approximately 0.45 m/s lower than that of the control group, with a decline as high as 32.8%. In addition, Genin et al. also found that the limit of walking speed of transfemoral amputees was merely of 1.2 m/s (Genin et al., 2008). In Chapter VI of this paper, the fast walking speed of amputees was only of approximately 1.1 m/s, which was dramatically lower than that of abled persons at nearly 2 m/s. The maximum walking speed that amputees can control is an indicator of the ability to control their movements. In this regard, it is a great functional improvement for amputees to improve the motion control ability and increase their walking speed.

It is suggested that more in-depth research should be carried out on the joint angular velocity of the amputee's gait. Clinical rehabilitation should also be combined with inter-joint coordination and DI as indicators, aiming at the lower limb sound joints of amputees with targeted exercises, so as to enhance their athletic ability. In the

past, rehabilitation training focused more on the athletic training of the amputated limbs. However, it can be seen that the non-amputated limb training is of equal importance from this study. Improving the athletic ability of the hip joints and ensuring the stability of the ankle joints in the non-amputated limbs will help to improve the gait coordination.

4.7 Summary

In this chapter, we analyzed the gait coordination of unilateral transfemoral amputees wearing traditional passive transfemoral prostheses. The gait coordination was represented by inter-joint coordination index, which was quantified by the CRP. The inter-joint coordination of the hip–knee, knee–ankle, and hip–ankle joint groups was studied. Moreover, the decomposition movement was also considered. Amputees and abled persons were compared to identify the differences.

Compared with the conventional gait analysis for transfemoral amputees, the gait coordination analysis in this paper can distinguish not only the kinematics changes single joint, but also the changes in the coordination relationship between joints. The advantage enables the gait deviation of amputees can be interpreted in a more in-depth way. In addition, the inter-joint coordination index depends on two variables, namely, joint angle and joint angular velocity. Thus, it has a special advantage in sensitively detecting the gait differences.

The insufficiency of the prosthesis in compensating the biomechanical function of amputees may lead to changes in the control strategy of other parts of the body, thus changing the coordination. In this case, amputees tend to walk using a coordination

pattern different from that of abled persons. However, even if the coordination pattern differs from that of abled persons, there will be consistency of features among amputees.

The prostheses involved in this experiment exerted similar impact on the gait coordination. Although different amputees wore different models of prostheses, there is little difference in gait coordination among amputees, which indicates similar reliability of the prostheses in satisfying the basic functions. In addition, it also proved that the amputees were skilled in walking and fully adapted to such prostheses, which are our inclusion criteria for the recruitment.

The hip joints and ankle joints of the amputees' non-amputated limbs played a significant role in the adjustment of gait coordination. The hip joint is responsible for shortening the duration of swing phase of the non-amputated limb which is beneficial for ensuring the balance of gait; while the ankle joint reduced the degree of freedom of motion control by prolonging the duration of the plantarflexion moment, which makes it easier to adjust the motion control. In addition, the adjustment of coordination related to ankle joints enables the metabolic expenditure of gait to be reduced. In the process of clinical rehabilitation, we may attempt to improve the athletic ability of the hip joints of amputees' non-amputated limbs, and use wearable accessories to assist the ankle joints of their non-amputated limbs, so as to ensure the stability and enhance their athletic abilities.

The knee–ankle inter-joint coordination of the prosthesis has shown significant differences from that of abled persons owing to its different mechanical mechanism, which also affects the gait coordination. In the future design of prostheses, this insufficient biomechanical function should be compensated so as to improve the

gait coordination.

In addition, our study also shows that there is still a huge space to improve the function of the prosthetic knee. The prosthetic knee reduces the degree of freedom of motion in the stance phase, which ensures the safety while exerting a negative effect. Other sound joints of the body need to make more adjustments to compensate for the missing motor function of the prosthetic knee joint. When the amputated limb is in the single support phase, the contralateral limb is in the swing phase. The inadequate ability of the amputated limb to control its motion makes it necessary for the amputees to shorten the single support phase. That is to say, the swing phase of the non-amputated limb on the opposite side is shortened and the limbs enter the double support phase earlier to ensure gait stability. In the future, the prosthesis design should compensate the shortage of this function so as to improve the gait coordination of amputees.

The slower walking speed of amputees is directly related to the adjustment of gait coordination. Owing to the limited ability of the amputees to control their motions, the motion will be slowed down. The decomposition movement is utilized to ensure the accuracy and stability of movement. These changes directly lead to the walking speed being lower than that of abled persons, which could be regarded as a compromise procedure.

The athletic ability of amputees' sound joints in the lower limbs needs to be strengthened urgently. More attention needs to be paid to the angular velocity of the amputees' lower limb joints in gait analysis. By reducing the impact on the angular velocity of the sound joint, and conducting targeted training on the joints, the amputees will be able to walk faster with higher angular velocity of the joints, which

will directly improve the gait coordination.

In general, the different coordination revealed that there are changes in the control strategies of amputees to cope with the insufficient biomechanical function of the musculoskeletal system, which is a countermeasure to the functional deficiencies. The prosthetic knee and ankle joints work in a completely different way from those of abled persons. The different joint function and the insufficient ability of controlling the prosthesis are the direct factors that lead to the change in motion control strategy. Therefore, the inter-joint coordination of the lower limbs varies accordingly. The study of the inter-joint coordination of transfemoral amputees will contribute to a further knowledge of amputees' motion control characteristics; what is more, to have a better understanding of the functional defects of the prostheses. In the clinic, the coordination analysis of the lower limb joints can also be used to conduct an effective evaluation for rehabilitation. It can serve as another criterion for gait performance. The programs of rehabilitation therapy could be improved accordingly, and hence, the gait coordination of amputees can be improved.

CHAPTER V EFFECTS OF THE KNEE–ANKLE COUPLING DESIGNED PROSTHESIS ON GAIT COORDINATION

According to the previous chapter, it has been discovered that the coordination analysis can intuitively detect the coordination deviation and sensitively embody the coupling variation between joints. Meanwhile, the knee–ankle coordination of the transfemoral prostheses was found to be substantially different from that of abled persons, and the defects of the prosthetic knee and ankle also affected gait coordination. This chapter presents the research conducted on the gait coordination of the knee–ankle coupling designed prosthesis and evaluated whether the knee–ankle coupling designed mechanism can effectively improve gait coordination of amputees.

Six amputees with transfemoral amputation participated in the experiment mentioned in this chapter. Amputees No. 7 and No. 8 (see Table 3-1 in Chapter III) were excluded from this experiment. Because the knee–ankle connectors (the prosthetic shank) of the designed knee–ankle coupling prosthesis is irreplaceable, the lower limbs' lengths of the two amputees failed to match the knee–ankle coupling designed prosthesis applied in this experiment. Other amputees underwent the gait analysis, A and B, which was referred to in Section 3.1, Chapter III. All the amputees participated in the gait experiments at the preferred speed with which they felt comfortable.

To facilitate the distinction, the gait of the prosthesis and non-amputated limb are respectively expressed as Prosthesis A and Non-amputated limb A in the gait

analysis where amputees wore the daily used transfemoral prostheses, and were expressed as Amputated limb B and Non-amputated limb B in the gait analysis where amputees wore the knee–ankle coupling designed prosthesis, as shown in Fig. 5-1.

（a）   （b）
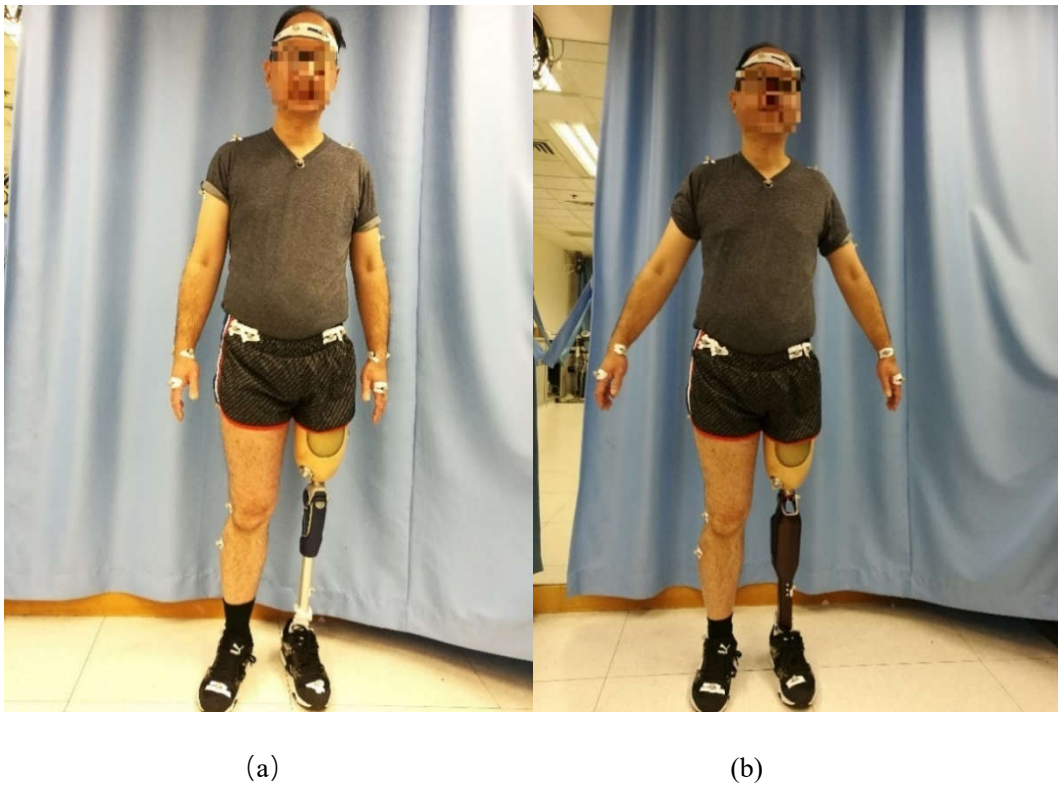
Fig. 5-1 Demonstration of amputees wearing two types of prosthesis: (a) traditional passive prosthesis; (b) knee–ankle coupling designed prosthesis

The gait cycle was categorized into phases for discussion (refer to Section 2.1, Chapter II for detailed definitions):

a) Stance phase: initial contact (IC), loading response (LR), middle stance (MSt), terminal stance (TSt), pre-swing (PS).

b) Swing phase: initial swing (IS), mid-swing (MSw), and terminal swing (TSw).

The gait differences of amputees who wear the two types of prostheses were analyzed statistically based on kinematics, dynamics, and coordination parameters. Due to that the purpose of the study in this chapter was to compare the differences of inter-joint coordination when wearing the two different types of prostheses and the differences between prosthetic gait and the gait of able-bodied persons had been discussed in Chapter IV, the results of control group were presented for reference without statistically analysis. In this chapter of study, the walking speed, maximum ground reaction force, lower limb joint angle, joint angular velocity, joint moment, distance from joint moment to rotation center, and joint coordination RMS, DP, and DI (see Section 3.3.1, Chapter III for details) were analyzed through an Independent-sample T-test to confirm whether there is significant difference. Separate discussions will be carried out in terms of the amputated limbs and non-amputated limbs. The significance level was set at $\alpha = 0.05$.

5.1 Results of gait analysis for the knee–ankle coupling designed transfemoral prosthesis

Based on the comparison of spatio-temporal parameters, there was no significant difference in the duration of gait cycle and walking speed when amputees wore the two types of prostheses, whatever in the amputated limb or non-amputated limb, as shown in Fig. 5-1. No significant difference in gait duration or walking speed was found between the bilateral limbs of amputees. The gait duration of amputees was approximately 0.17 s longer than that of the control group, and the walking speed was approximately 0.34 m/s slower than that of the control group. However, there was significant difference of stride length of the amputated limb when wearing the two types of prostheses (p = 0.049). On average, the stride length of the knee–ankle

coupling designed prosthesis (1.32 m) was 0.13 m larger than that of the traditional passive prosthesis (1.19 m). When wearing the two types of prostheses, there was no significant difference in the time proportion of the stance and swing phases, whatever in the amputated limb or non-amputated limb. The time proportion of the stance phase was approximately 58% in amputated limb, and that of the non-amputated limb was approximately 65%. However, it was very different from that of abled persons (approximately 61%), as shown in Fig. 5-2.
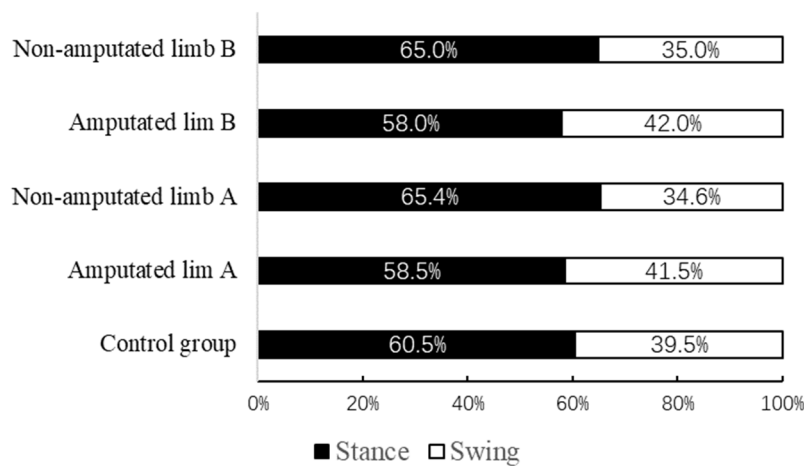
Fig. 5-2 Duration ratio of stance and swing phases for amputees wearing two types of prostheses and compared with that of the control group.

As shown in Fig. 5-3, the joint angle takes the neutral position as 0, positive hip and knee joint angles represent the joint flexion, and the ankle joint angles are positive for the plantarflexion, vice versa;. hip and knee moments are positive for the extension moment, and positive ankle moments represent the plantarflexion moment, vice versa. Values of the GRF are positive for the anterior direction and negative for the posterior one. The above mentioned positive and negative values only indicate the direction.

Table 5-1 Comparison of spatio-temporal, kinematics, and kinetic parameters for amputated and non-amputated limbs when wearing two types of prostheses, and limbs of the control group, respectively.

Items	Parameters	Control group	Amputees with prosthesis A		Amputees with Prosthesis B	
			Amputated limbs A	Non-amputated limbs A	Amputated limbs B	Non-amputated limbs B
Spatio-temporal	Gait duration (s)	1.11 ± 0.04	1.30 ± 0.10	1.27 ± 0.09	1.28 ± 0.09	1.28 ± 0.09
	Walking speed (m/s)	1.37 ± 0.13	0.94 ± 0.06	0.94 ± 0.07	1.03 ± 0.13	1.04 ± 0.11
	Stride length (m)	1.37 ± 0.13	1.19 ± 0.04	1.23 ± 0.06	1.32 ± 0.12★	1.33 ± 0.10
Kinematics	Hip angle at HS (°)	28.80 ± 5.30	29.70 ± 4.60	28.40 ± 6.90	27.80 ± 6.00	30.9 ± 7.9
	Max hip extension angle (°)	−11.4 ± 4.30	−10.00 ± 2.30	−11.90 ± 4.50	−13.60 ± 2.90★	−14.2 ± 4.5
	Max knee flexion angle (°)	60.7 0± 8.80	57.00 ± 11.20	58.80 ± 8.90	62.00 ± 8.20	50.2 ± 14.3
	Max plantarflexion angle in LR (°)	−2.00 ± 2.80	−4.50 ± 2.50	−6.90 ± 3.40	−6.10 ± 2.80	−6.6 ± 4.10
	Max dorsiflexion in stance phase (°)	10.70 ± 3.2	10.70 ± 5.80	4.50 ± 5.60	4.50 ± 5.60	1.30 ± 4.00
Kinetics	Max hip flexion moment (Nm/BW)	−0.95 ± 0.47	−0.67 ± 0.19	−0.35 ± 0.22	−0.71 ± 0.19	−0.35 ± 0.20
	Max knee extension moment in MSt (Nm/BW)	0.52 ± 0.30	0.18 ± 0.22	0.00 ± 0.25	0.00 ± 0.10	0.04 ± 0.30
	Max knee flexion moment in TSt (Nm/BW)	−0.14 ± 0.23	−0.26 ± 0.17	−0.50 ± 0.32	−0.54 ± 0.12★	−0.56 ± 0.30

	Max plantarflexion moment (Nm/BW)	1.37 ± 0.21	1.18 ± 0.23	1.32 ± 0.18	1.14 ± 0.13	1.39 ± 0.30
	Max vertical GRF (N/BW)	10.80 ± 0.80	10.00 ± 0.50	10.40 ± 0.60	9.70 ± 0.90	10.40 ± 0.60
	Max anterior GRF (N/BW)	−1.60 ± 0.36	−0.36 ± 0.24	−0.90 ± 0.31	−0.40 ± 0.38	−0.85 ± 0.42
	Max posterior GRF (N/BW)	1.94 ± 0.40	0.38 ± 0.40	0.37 ± 0.72	0.45 ± 0.43	0.36 ± 0.76

HS: heel strike; LR: loading response; BW: body weight; ★ indicates that there is a significant difference between the corresponding parameters when amputees wear Prosthesis A and Prosthesis limb B ($p < 0.05$).

There were differences in joint angle and joint moment of the lower limbs when wearing the two types of prostheses. Although there was no significant difference in the hip joint angle of the amputated limb at the beginning of the gait cycle when amputees wore the two types of prostheses, the maximum hip extension angle of the amputated limb (occurred at approximately 53% gait cycle) when wearing the knee–ankle joint prosthesis was approximately 3.6° higher, compared to the traditional prostheses ($-13.6° \pm 2.9°$ vs. $-10° \pm 2.3°$, p=0.04), as shown in Fig. 5-3 a. A further study on the joint angular velocity found that at the TSt (approximately 48% gait cycle), the hip angular velocity of the limb wearing knee-ankle coupling designed prosthesis was significantly higher than that with traditional passive prostheses, with the maximum difference of 21°/s (P = 0.04). Even so, the hip moment of the amputated limb when wearing the two types of prosthesis presented a small difference, and the joint angle and moment did not have a significant difference in the whole gait cycle.

(a)

(b)

(c)

(d)
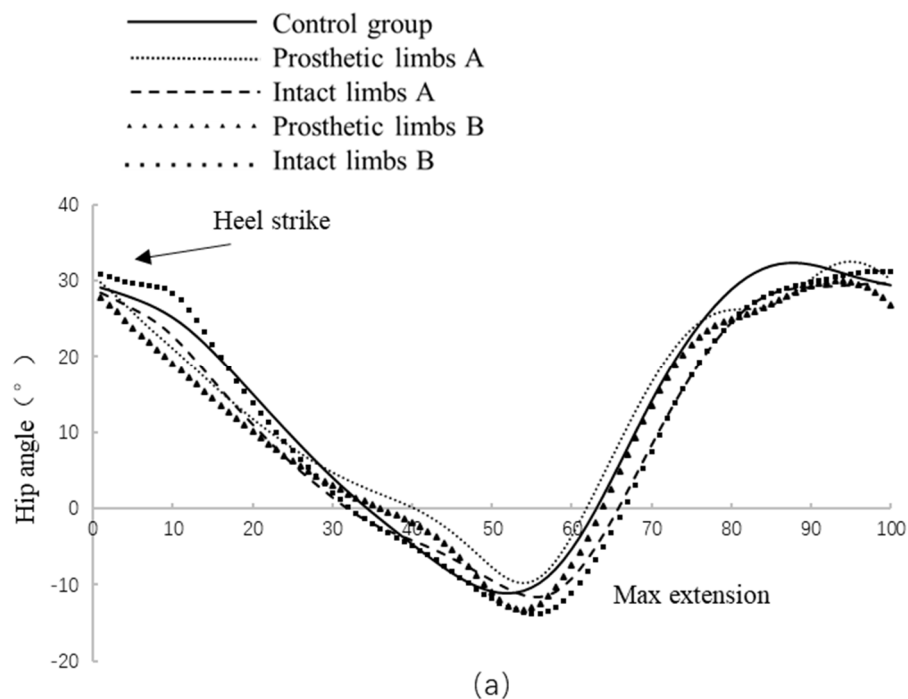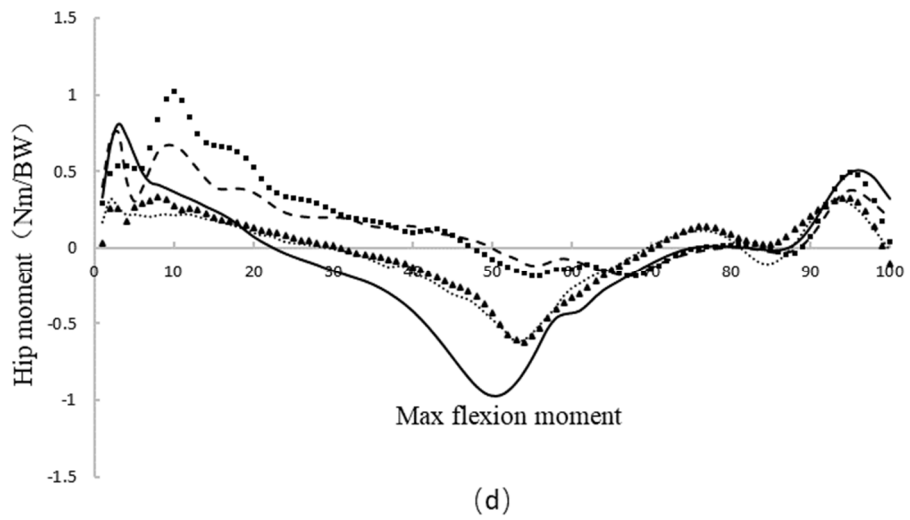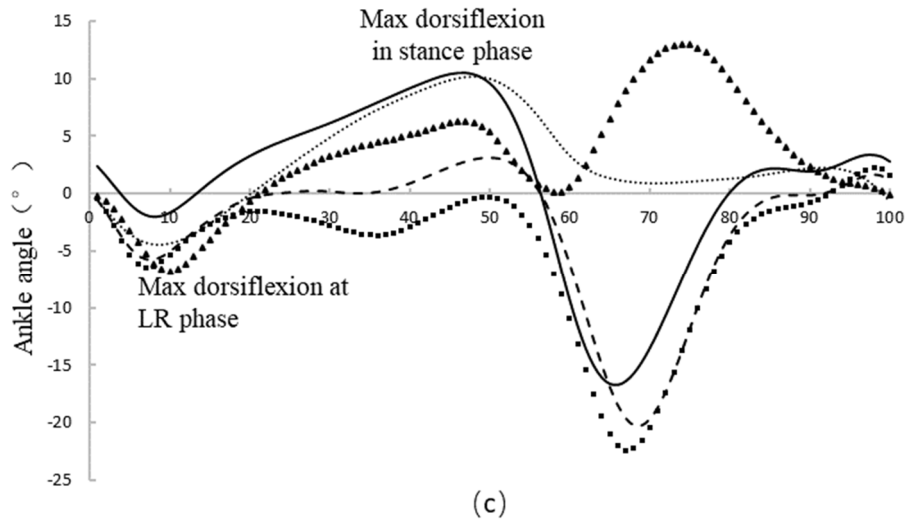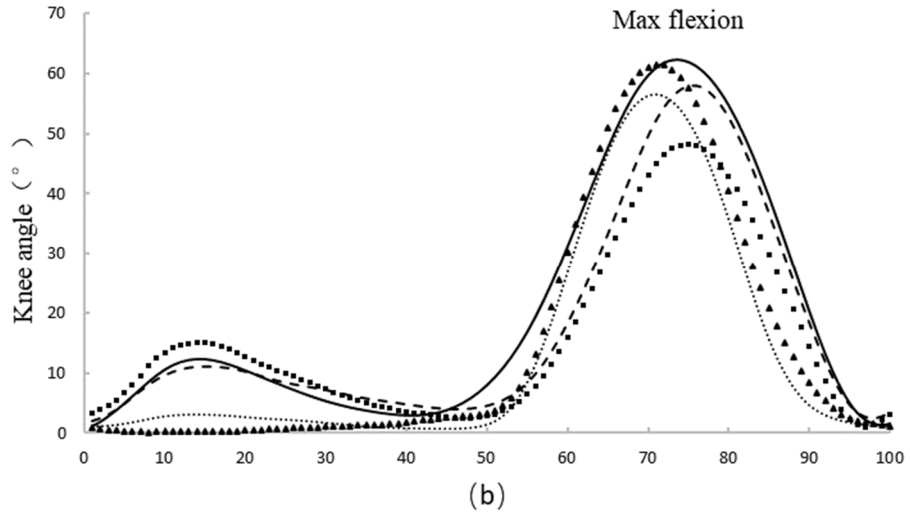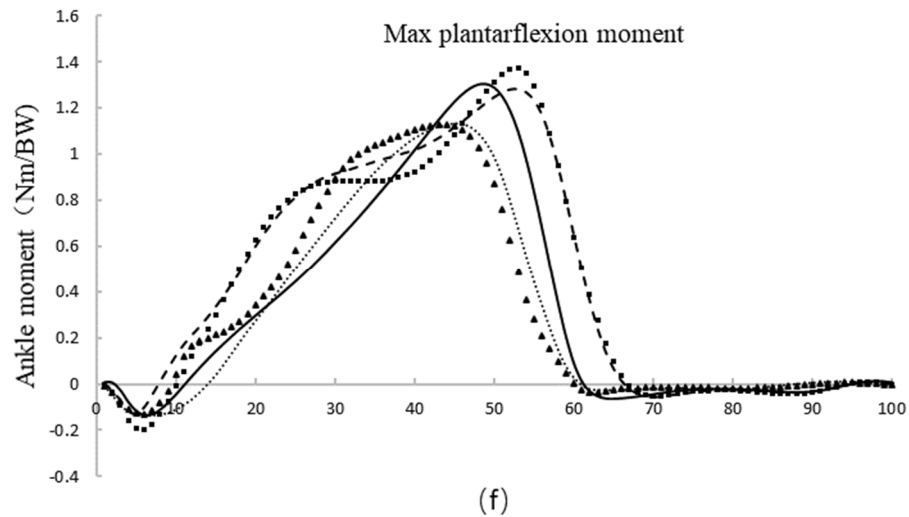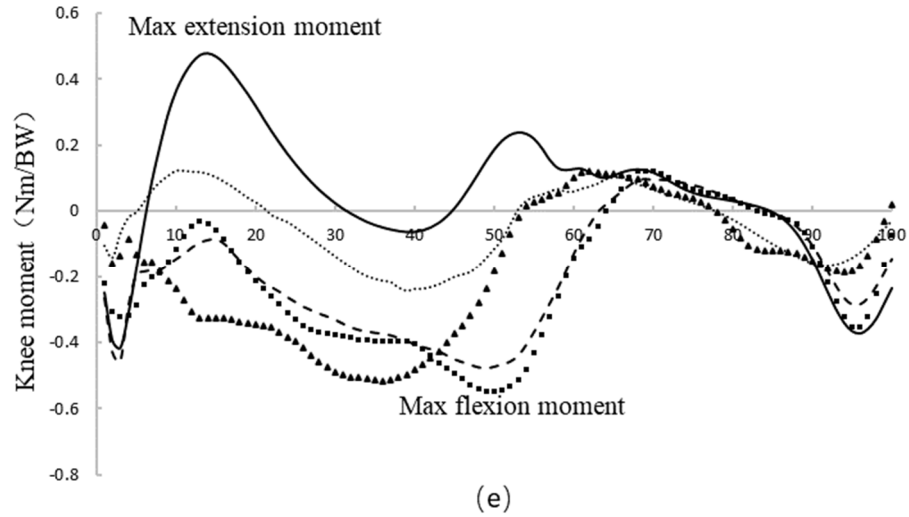
Fig. 5-3 Joint angle and joint moment in 100% gait cycle: (a) hip joint angle; (b) knee joint angle; (c) ankle joint angle; (d) hip moment; (e) knee moment; (f) ankle moment

The prosthetic knee joint angles showed no significant difference when wearing the two types of prostheses in the whole gait cycle, but the joint moments were quite different, as shown in Fig. 5-3 e. From 7% gait cycle to 43% gait cycle, the knee moment of the knee–ankle coupling designed prosthesis was significantly higher

than that of the traditional prosthesis. The peak knee flexion moment of the knee–ankle coupling designed prosthesis was −0.54 ± 0.12, approximately 107.7% higher than that of the traditional prosthetic knee (p < 0.01). Although the prosthetic knee joints were very different in the mechanism, they did not induce a significant difference in the knee joint of non-amputated limbs, the joint angle and joint moment of the non-amputated limbs were similar when wearing the two types of prostheses.

The prosthetic ankle joint angles in the two types of prosthetic gaits were not significantly different in the stance phase. However, an in-depth research on the ankle joint angular velocity proved that the velocity of the knee–ankle coupling designed prosthesis was significantly higher than that of the traditional prosthesis in the LR phase after heel strike, with the maximum difference reaching 58°/s (p = 0.02).

(a)
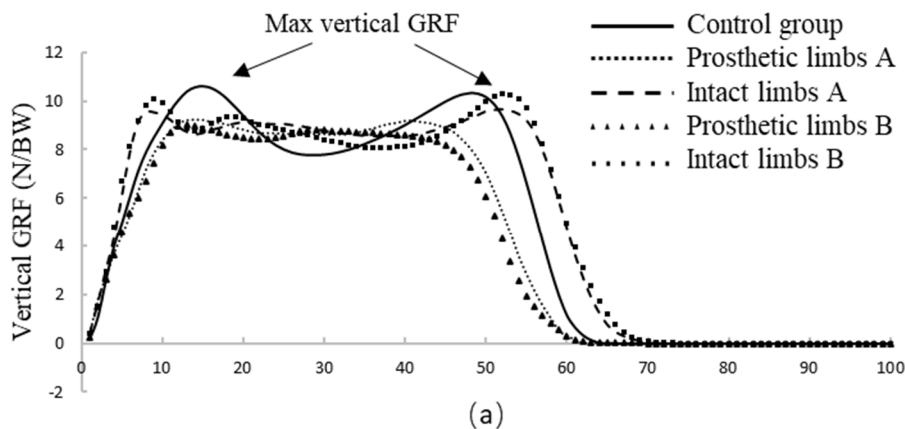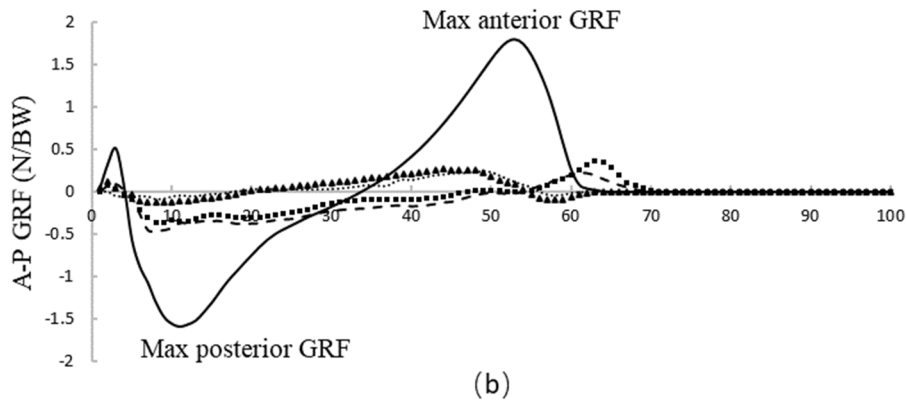
Fig. 5-4 Ground reaction force (GRF) of amputated and non-amputated limbs of amputees wearing the two types of prostheses, and compared with that of the control group: (a) vertical GRF; (b) anterior–posterior GRF.

During the whole gait cycle, wearing the knee-ankle coupling designed prosthesis did not significantly change the GRF in the amputees' gait (Fig. 5-4). The GRF in the vertical and the anterior–posterior direction when wearing the two types of prostheses were similar, and the maximum GRF towards the vertical and the anterior–posterior direction behaved with no significant difference (Table 5-1).

The inter-joint coordination of the lower limbs of the subjects is shown in Fig. 5-5, where 0° CRP means that the movements of the two joints are completely synchronized, the positive values indicate that the proximal joint movement is ahead of the distal joint and vice versa, and ± 180° CRP means that the two joints move in inverse phase.
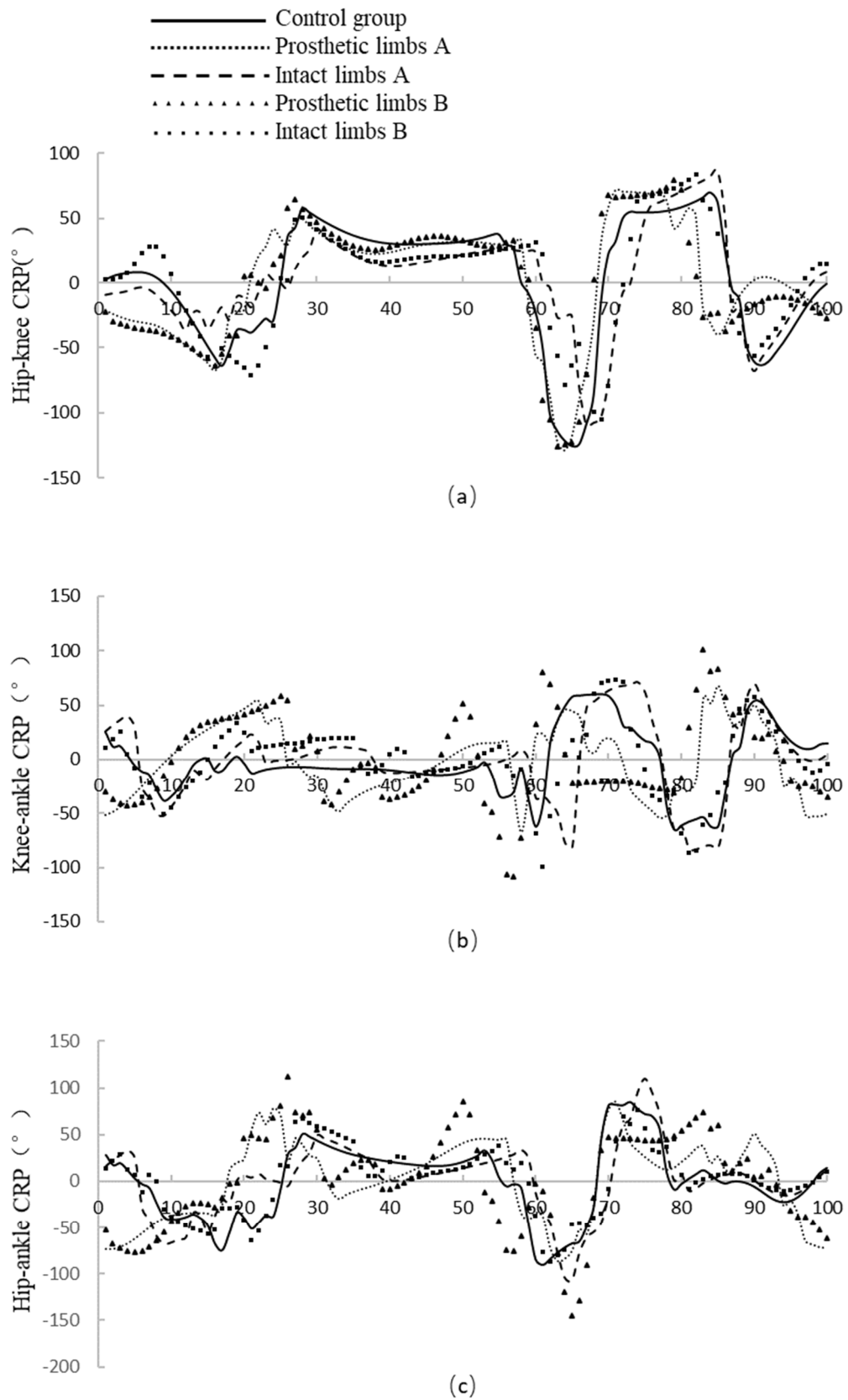
Fig. 5-5 Ensemble mean CRP profile of amputated and non-amputated limbs, and

limbs of the control group: (a) hip–knee; (b) knee–ankle; (c) hip–ankle

It was found from the CRP profile that wearing the knee–ankle coupling designed prosthesis resulted in a small change in hip–knee coordination for both the lower limbs. There was no significant difference in hip–knee coordination when wearing the two types of prostheses for both lower limbs, as illustrated in Fig. 5-5 a. Similarly, no significant change occurred in the knee–ankle and hip–ankle coordination of non-amputated limbs after wearing the knee–ankle coupling designed prosthesis. However, the coordination of the knee–ankle and hip–ankle of amputated limbs was significantly different when wearing the two types of prostheses. From TSt to IS (approximately 49% gait cycle to 72% gait cycle), the amputated limb knee–ankle and hip–ankle coordination experienced significant difference, as shown in Fig. 5-5 b and Fig. 5-5 c.

Table 5-2 CC of the CRP between amputees and control group.

Joint groups	Gait phase	Amputated limbs A	Non-amputated limbs A	Amputated limbs B	Non-amputated limbs B
Hip–knee	Stance	0.76	0.86	0.79	0.89
	Swing	0.81	0.93	0.79	0.90
Knee–ankle	Stance	0.04	0.80	0.41	0.48
	Swing	0.32	0.85	−0.09	0.75
Hip–ankle	Stance	0.24	0.78	0.48	0.81
	Swing	0.63	0.89	0.39	0.85

The inter-joint coordination pattern is illustrated by both the CC and RMS. It was found that the hip–knee coordination patterns were similar when wearing the two types of prostheses for both amputated and non-amputated limbs (Tables 5-2 and 5-3). The difference in CC value (compared with that of the control group) of the hip–knee coordination was less than 0.03 for both amputated and non-amputated limbs

while wearing the two types of prostheses. Similarly, the RMS showed no significant difference in hip–knee coordination under the two conditions for both amputated and non-amputated limbs. Wearing the knee–ankle coupling designed prosthesis did not make the knee–ankle coordination of amputees more similar to that of the control group.

Table 5-3 RMS of the CRP for amputated and non-amputated limbs when wearing two types of prostheses, and limbs of the control group, respectively.

Joint groups	Gait phase	Control groups	Amputated limbs A	Non-amputated limbs A	Amputated limbs B	Non-amputated limbs B
Hip-knee	Stance	44.8 ± 4.4	43.5 ± 3.0	44.3 ± 4.4	46.2 ± 4.3	47.7 ± 4.8
	Swing	70.4 ± 6.8	70.3 ± 7.5	73.9 ± 6.5	68.5 ± 2.6	72.4 ± 10
Knee-ankle	Stance	33.4 ± 5.8	41.2 ± 7.2	43.0 ± 7.1	50.9 ± 7.4★	43.4 ± 9.4
	Swing	56.1 ± 7.7	60.0 ± 14.8	63.9 ± 5.0	62.4 ± 6.4	65.2 ± 3.5
Hip-ankle	Stance	46.2 ± 10.1	57.5 ± 7.4	54.40 ± 4	63.0 ± 8.8	55.5 ± 4.6
	Swing	53.8 ± 14.9	66.4 ± 8.3	59 ± 10.1	69.3 ± 9.3	57.2 ± 7.7

The CC of knee–ankle coordination was both less than 0.5 at the two conditions in the amputated limbs. The RMS of the knee–ankle of amputated limbs in the stance phase significantly changed, accounting for 50.9° ± 7.4°, compared with that when wearing the traditional prostheses. The change makes it more different from that of the control group. While wearing the knee–ankle coupling designed prosthesis, the knee–ankle coordination of non-amputated limbs was less close to that of the control group, with the CC value amounting to only 0.48 in the stance phase and a decrease of 0.1 in the swing phase. However, the RMS values of the knee–ankle CRP of non-amputated limbs showed no significant difference when wearing the two types of prostheses.

In terms of hip–ankle coordination on the amputated limb, wearing the knee–ankle coupling designed prosthesis did not result in a greater similarity with that of abled persons. Compared with the control group, the CC value of hip–ankle coordination of amputated limbs when wearing the knee–ankle coupling designed prosthesis was less than 0.5 in both the stance and swing phases. There was no significant difference in the hip–ankle RMS of amputated limbs between the two types of prostheses. The CC value of hip–ankle coordination of non-amputated limbs varied little after wearing the knee–ankle coupling designed prosthesis, and it increased by 0.03 and reduced by 0.04 in the stance and swing phases, respectively. No difference occurred in hip–ankle RMS in both the stance and swing phases when wearing the two types of prostheses.

(a)

(b)
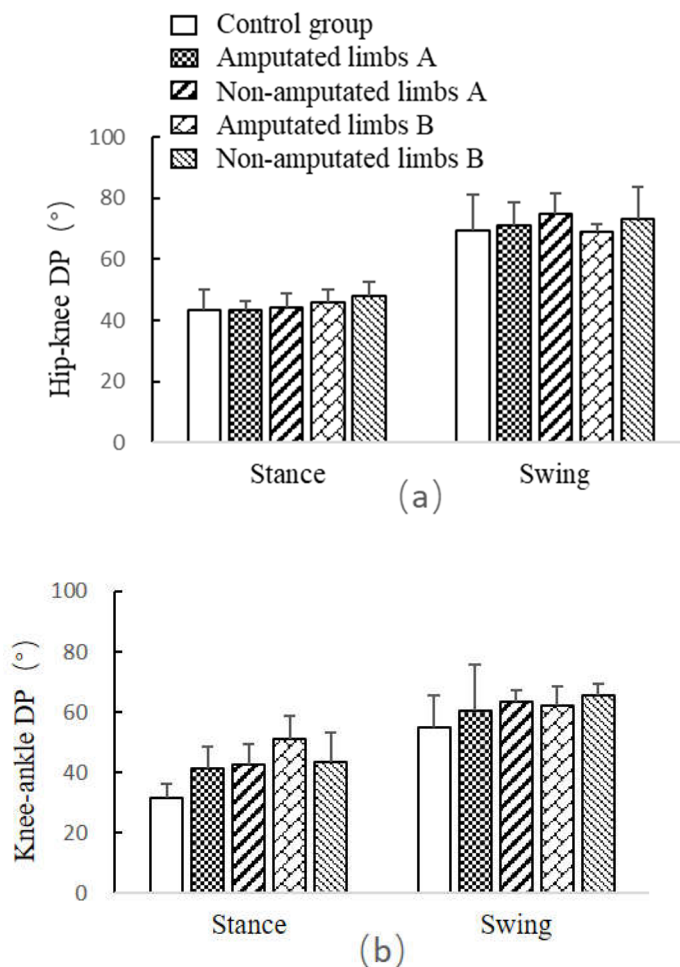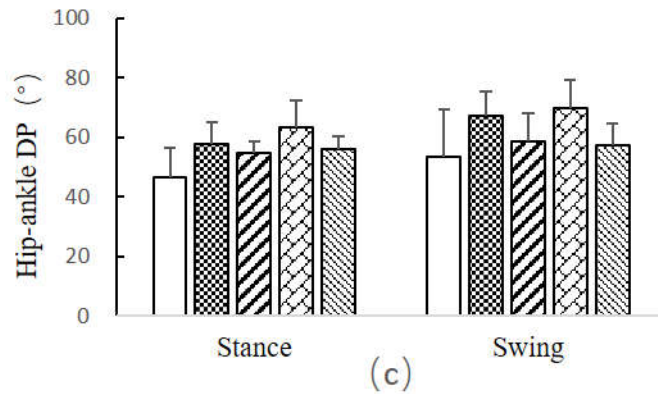
Fig. 5-6 DP for amputated and non-amputated limbs when wearing two types of prostheses, and limbs of the control group: (a) hip–knee; (b) knee–ankle; (c) hip–ankle. There was no significant difference in the DP value of each joint group under the two conditions (p > 0.05).

Judging from the variation in joint coordination among amputees, the uniform knee–ankle coupling designed prosthesis failed to reduce the degree of variation (p > 0.05) compared to that of disunity prostheses, as shown in Fig. 5-6. There was no significant difference in hip–knee, knee–ankle, and hip–ankle coordination between the two gait conditions for both the amputated and non-amputated limbs, whatever the phase, stance or swing.

According to the statistical results of DI, the prosthetic ankle joint related DI changed significantly owing to the dorsiflexion movement in the swing phase, but developed no significant difference in other joint groups of both lower limbs after the amputee wore the knee–ankle coupling designed prosthesis, as shown in Fig. 5-7. The knee–ankle and hip–ankle DI in the amputated limb suffered from a significant reduction in the swing phase. Knee-ankle DI changed from 40% ± 31.9% of the traditional passive transfemoral prosthesis to 9.2% ± 6.9% of the knee–ankle coupling designed prosthesis, and it was closer to the 7.4% ± 4.8% of the control

group; the hip–ankle DI after wearing the knee–ankle coupling designed prosthesis decreased from 39.3% ± 31.5% with the traditional passive transfemoral prosthesis to 9.2% ± 5.7%, closer to the 9.6% ± 6.8% of the control group.
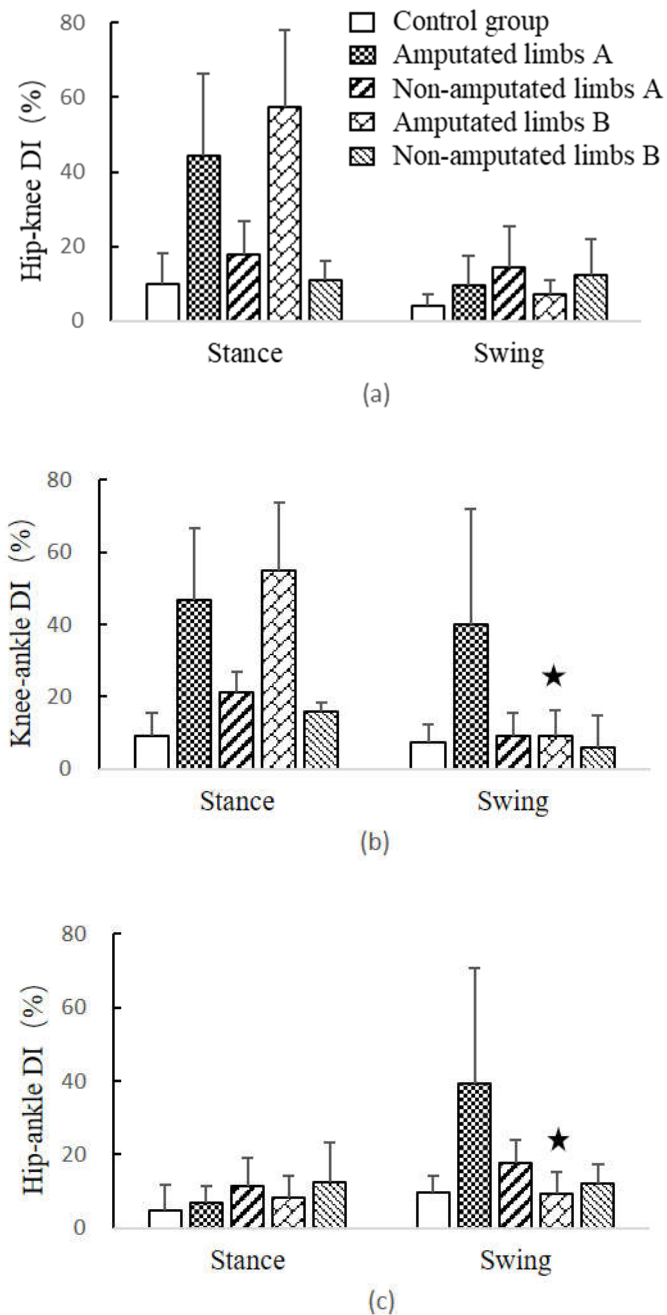
Fig. 5-7 DI for amputated and non-amputated limbs when wearing two types of prostheses, and limbs of the control group: (a) hip–knee; (b) knee–ankle; (c) hip–ankle; ★ represents that there was a significant difference when wearing the two

types of prostheses (p < 0.05).

This analysis presented in this chapter aims at validating whether the knee–ankle coupling designed prosthesis can improve amputees' gait coordination effectively. We compared the amputees' inter-joint coordination of lower limbs when wearing the traditional passive prostheses and the knee–ankle coupling designed prosthesis and analyzed the changes in the gait. The advantages of the knee–ankle coupling designed mechanism and the deficiencies that need to be improved in the future will be investigated.

5.2 Features of the knee–ankle coupling designed mechanism

The biggest difference between the knee–ankle coupling designed prosthesis and the traditional passive prosthesis lies in the ankle dorsiflexion and plantarflexion function. The knee-ankle coupling designed prosthesis has the advantages in "active" rotating the ankle joint, which are exactly what the traditional passive transfemoral prosthesis lacks. As a result, the knee-ankle coupling designed mechanism of the ankle joint is quite different from that of the traditional prosthesis. Through the comparison of this coupling designed prosthesis with single-axis knee joint prostheses, Sapin et al. revealed that the knee–ankle coupling designed prosthesis have the following advantages (Sapin et al., 2008): Firstly, it could enable the sole of the foot to touch the ground faster at the beginning of a gait cycle; secondly, it was feasible to increase the toe height off the ground by means of the dorsiflexion of the prosthetic ankle in the swing phase; what is more, the moment of the knee joint was greater when the knee joints were extended in the stance phase. All these three features can make the gait safer and more stable. For a validation, the analysis performed in this study obtained results consistent with those of Spain et al. In

addition, we also analyzed the gait coordination by combining spatio-temporal and kinetics parameters and found more differences while wearing the knee–ankle coupling designed prostheses.

From the spatio-temporal parameters (Table 5-1), there was no significant difference in the duration of the gait cycle and walking speed of amputees when wearing the two types of prostheses. The gait duration was approximately 0.17 s longer and the walking speed was approximately 0.34 m/s slower than that of the control group, which indicates that amputees still tend to walk in a slower speed. There was also no significant difference between the two conditions in the duration proportion of the stance and swing phases in the whole gait cycle. The proportion of the stance phase of the amputated limbs and non-amputated limbs was still approximately 4.7% higher and 2.3% lower than that of the control group, respectively. Even though the above parameters were not significantly improved, the amputated limbs' gait stride was significantly higher (approximately 0.13 m) when wearing the designed knee–ankle coupling prosthesis than that of the traditional prosthesis.

5.3 Effects of the knee–ankle coupling designed mechanism on gait coordination

Except for the prosthetic knee–ankle coordination, the other four groups of inter-joint coordination patterns were similar when wearing the two types of prostheses. The knee–ankle coupling designed prosthesis still maintains a basic function similar to that of other passive transfemoral prostheses and has a relatively consistent influence on the coordination of other sound joints.

The most significant difference in inter-joint coordination between the two walking conditions occurred before the swing phase. The knee–ankle coupling designed prosthesis is able to dorsiflex actively in the swing phase. The dorsiflexion movement started at the PS phase of gait, namely, before the end of the stance phase and was willing to enter the swing phase, as shown in Fig. 5-3 c. In fact, the height of the prosthetic toes from the ground has always been an important index for prosthesis design and clinical rehabilitation. If the vertical clearance is not sufficient, the amputees' prostheses may touch the ground when swinging the limb, thus causing problems such as falls. This knee–ankle coupling designed prosthesis greatly increased the height of the amputees' toes off the ground and improved the safety. The increase in clearance height can effectively prevent the occurrence of pathological gaits such as the circumducted, abducted, or vaulting gait. When the ankle dorsiflexed, there was no significant difference between the angle of the prosthetic knee and the traditional prosthetic knee. After that, the prosthetic hip reached the maximum extension angle at approximately 53% gait cycle. During this period, the lower limbs continued to rotate forward. The knee angle of the knee–ankle coupling designed prosthesis did not change compared with that of the traditional prosthesis, and the thigh rotated forward together with the shank. The ankle dorsiflexion of knee–ankle coupling designed prosthesis increases the rotation angle of the thigh compared with that when wearing the traditional prosthesis, resulting in that the maximum extension angle of the hip joint increased by approximately 3.6°. The lower limbs rotate forward like an inverted pendulum, and the increase in the hip maximum extension angle makes the lower limb rotate forward more anterior (Kuo, 2007). The support foot shifts the center of body mass forward with the inertia, thus causing an increase of 0.13 m in the stride length. In

this process, the dorsiflexion of the prosthetic ankle joint ensures compliance of lower limb rotation. Meanwhile, the change in hip and ankle angles of the amputated limb when wearing the knee–ankle coupling designed prosthesis also makes the hip–ankle CRP different from when wearing the traditional prosthesis, as shown in Fig. 5-3 c.

Although the stride length of the amputated limb's gait significantly increased when wearing the knee–ankle coupling designed prosthesis, the maximum anterior–posterior GRF exerted no significant change (Table 5-1). Wearing the knee–ankle coupling designed prosthesis did not improve the defect of insufficient propulsion force for amputees. The increase in stride length is only caused by the change in hip and ankle angles, and the center of body mass shifted forward due to inertia. Therefore, this change was not accompanied by changes in kinetics, such as joint moments of hip and ankle. For this reason, although the stride length was increased, the durations of gait cycle, walking speed, and GRF were not changed significantly.

Amputees wore different models of traditional prostheses, but the knee–ankle coupling designed prostheses were unified (Hydracandence knee "1P50", refer to Fig. 1-5). In spite of this, there was no significant change in the variation of inter-joint coordination when wearing the two types of prostheses. As stated in Chapter IV, our research on traditional prostheses had revealed that the inter-joint coordination variation was similar between the amputees group and control group, except the inter-joint coordination related to the prosthetic ankle. Even if different types of prostheses were involved, the coordination variation among the amputees was still within the normal range. As discussed in this chapter, amputees who wearing unified prostheses did not reduce the inter-joint coordination variation

significantly, as shown in Fig. 5-6.

Research on the DI found that the knee–ankle coupling designed prosthesis scarcely affects the control strategies of other joint groups, except the significantly reduced DI caused by the mechanism of ankle dorsiflexion in the swing phase (knee–ankle and hip–ankle in the amputated limb). The knee–ankle coupling designed function makes the movement of the prosthetic knee and ankle more similar to abled persons' musculoskeletal system, which was proved by the DI. By contrast, the knee–ankle coupling designed prosthesis showed little improvement in the decomposition movement of the sound joints. Namely, there was no significant change in the hip–knee DI in amputated limbs and in all the three joint groups of non-amputated limbs after wearing the knee–ankle coupling designed prosthesis.

5.4 Limitation of the current linkage mechanism

The knee–ankle coupling designed prosthesis also belongs to passive prosthesis, and thus, it maintains numerous similarities as other traditional prostheses involved in this study. On the one hand, knee–ankle coupling designed prosthesis inherited the common biomechanical functions of passive prosthesis, so that the DP and DI parameters in the two prosthetic gaits have similarities. On the other hand, the knee–ankle coupling designed mechanism plays a special role in improving gait coordination.

The defects of passive transfemoral prosthesis were discussed in the last chapter of this paper. One of the defects is that the reduction of the freedom of motion of the prosthetic knee in the stance phase limited the athletic ability of amputees. Although it can ensure walking safety, it affected the control strategies of other parts of the

body. The knee–ankle coupling designed prosthesis fails to amend this important functional deficiency. In the future prosthesis design, it can be considered to enhance the flexion function of the knee joint in the stance phase, of course, on the premise of ensuring a stable and safe gait. The prosthetic knee flexion directly affects the coordination of the ipsilateral hip joint, further affect the coordination of the joints of non-amputated limbs, and finally achieve an effect on the overall gait coordination of amputees.

The knee–ankle coupling designed function compensated traditional passive prosthesis' functional defect of relative independence working mechanism of the prosthetic knee and ankle. So that there was a certain coordination between the two prosthetic joints, which is closer to the musculoskeletal system of abled persons. This function enhances the prosthetic ankle function. It helps to dorsiflex the prosthetic ankle in the pre-swing and swing phases and increases the height of the toe off the ground to enhance gait safety. Besides, the improvement of the prosthetic ankle also helps to improve the compliance of hip joint extension. The improvement in biomechanical function originated from the driving effect of the knee joint on the ankle joint. Therefore, the knee–ankle coupling designed function can effectively improve the gait coordination of amputees, and the linkage mechanism has the advantages that other traditional artificial limbs.

However, there is also great room for improving the linkage mechanism. The knee joint of the current knee–ankle coupling designed prosthesis still maintains similar characteristics to those of other traditional prostheses. The linkage mechanism is mainly related to the driving effect of the knee joint on the ankle joint, and the influence of the ankle joint on the knee joint is not embodied. Therefore, the

displayed functional advantage lies mainly on the ankle joint. In the future design, on the basis of ensuring the functional advantages of the existing ankle joint, it is necessary to improve the prosthetic knee function through enhancing the feedback and adjustment from the ankle joint to the knee joint. For example, the prosthetic knee joint damping can be adjusted through the force feedback of the ankle joint in the stance phase to enhance the flexion function. Thus, gait coordination could be further improved.

5.5 Summary

This chapter analyzed the gait coordination of amputees wearing the knee–ankle coupling designed prosthesis and compared it with that when wearing traditional passive prostheses. The gait analysis was conducted with unilateral transfemoral amputees at a self-selected preferred speed which them feel comfortable. The aim of this study was to validate whether the linkage mechanism of the knee and ankle joints can effectively improve the gait coordination.

We confirmed that the knee–ankle coupling designed mechanism has certain advantages for improving gait coordination. In addition to the advantages of the linkage mechanism found by previous studies, this study also found that ankle dorsiflexion starting before the swing phase contributes to an improvement in amputees' gait coordination. Compared with the gait of wearing traditional prosthesis, the maximum hip extension angle of the amputated limb when wearing the knee-ankle coupling designed prosthesis increased. The limb rotated like an inverted pendulum into the swing phase with a larger amplitude, and the dorsiflexion movement of the ankle joint ensured the compliance of the rotation. These features significantly increased the stride length of the amputees' amputated

limb and improved their athletic ability.

Besides the above-mentioned improvements, the gait coordination when wearing the two types of prostheses was similar. The gait of the knee–ankle coupling designed prosthesis embodies many characteristics similar to those of the other passive transfemoral prostheses. Even though the models of traditional prostheses involved in this experiment were not unified, there was no significant difference in inter-joint coordination compared to the uniform knee–ankle coupling designed prosthesis. This can also be considered as an advantage that the special mechanism did not induce a negative effect on gait coordination. Because the knee-ankle coupling designed prosthesis maintained similar basic functions as other passive prostheses while also presented advantages in the inter-joint coordination of prosthetic knee and prosthetic ankle.

In general, the knee–ankle coupling designed mechanism is closer to the musculoskeletal system of abled persons than the knee–ankle split type prosthesis. Hence, it exhibits advantages in improving the gait coordination. However, there is still great room for improving the linkage function. The current knee–ankle coupling designed mechanism relies mainly on the ankle dorsiflex or plantarflex actively, which is driving by the prosthetic knee. The mechanism is dramatically different from the traditional prosthetic ankle joint. However, the knee joint is still relatively independent. The prosthetic knee maintains characteristics similar as those of the traditional passive prosthetic knee joint. In the future design, the function of the prosthetic knee needs to be enhanced, especially the active extension function in the stance phase. In this way the knee–ankle coupling designed mechanism will bring more positive changes to the gait coordination of amputees.

CHAPTER VI EFFECTS OF WALKING SPEED ON GAIT COORDINATION OF AMPUTEES

In this chapter, the gait coordination of transfemoral amputees is studied for different walking speeds. Subjects walked at three speeds in the experiment: the preferred, slow, and fast speeds. Gait coordination analysis of amputees is conducive to understanding the amputee's motion control strategies when walking at different speeds. It was hypothesized that the outcomes could identify the motion defects of amputees and guide the improvements in prosthesis design and clinical rehabilitation programs.

Although it would be easier to control the walking speeds by walking on a treadmill, it had been evidenced in other study that there were differences in lower limb inter-joint coordination between overground walking and treadmill walking (Chiu et al., 2015). What is more, the athletic ability of each amputees may not be the same. If we require everyone walk at a very strict speed condition, the gait would not be natural. Considering of this, we instructed the amputees to walk at the lowest speed of which them would adapt in their daily lives to be the slow speed; while walk at the fastest speed on the condition of a stable gait to be the fast speed.

A total of seven amputees completed the experiment successfully. Of the eight recruited amputees (refer to Table 3-1 and Table 3-2), amputee No. 3, failed to complete the experiment owing to the length of the strides, larger the measurable range of the force platform, and which were caused by a greater height. The amputees participated in gait analysis A (refer to Section 3.1.3, Chapter III for details).

The preferred speed means the speed at which amputees feel comfortable. Low and high speeds mean that amputees walk as slow and fast as possible, respectively. Amputees in the gait analysis will choose the walking speed randomly.

The paired T-test was performed to compare the differences when walking at different speeds. The parameters including walking speed, stride length, gait duration, time ratio of gait phase, joint angle, joint moment, GRF, and the coordination parameters RMS, DP, and DI were investigated. The slow speed and fast speed were compared to preferred speed, respectively. Separate discussions will be carried out in terms of the stance phase and swing phase of gaits. The significance level is set at $\alpha = 0.05$.

6.1 Results of gait analysis under three different speeds

Amputees behaved similarly in terms of the gait speed, stride length, and gait duration under different speeds for the amputated and non-amputated limbs, (the p-value of paired T-test samples were all greater than 0.05), as presented in Table 6-1. The walking speed, stride length, and gait duration of both lower limbs under the fast and slow speeds were significantly different from those at the preferred speed. The preferred speed was approximately 0.91 m/s. By contrast, the slow speed on average was approximately 15% lower (0.77 ± 0.12 m/s, and the p values were less than 0.01 for both lower limbs), while the fast speed was about 20% higher (1.09 ± 0.14 m/s, and the p values were less than 0.01 for both lower limbs, respectively). The gait duration was approximately 1.3 s at the preferred speed and about 10% higher and 10% lower at the slow and fast speeds, respectively. It was 1.44 ± 0.15 s under the slow speed, and the p values of the lower limbs in both sides were less than 0.01; while it was 1.16 ± 0.13 s under the fast speed, and the p values of the

lower limbs in both sides were less than 0.01. At the preferred speed, the stride length was approximately 1.19 m, whereas at the slow and fast gaits they were about 7% lower and 7% higher, respectively (it was 1.11 ± 0.09 m/s under the slow speed, and the p values of the amputated and non-amputated limb gaits were 0.04 and 0.06, respectively; while it was 1.27 ± 0.08 m/s under the fast speed, and the p values of amputated and non-amputated limbs were 0.049 and 0.04, respectively).
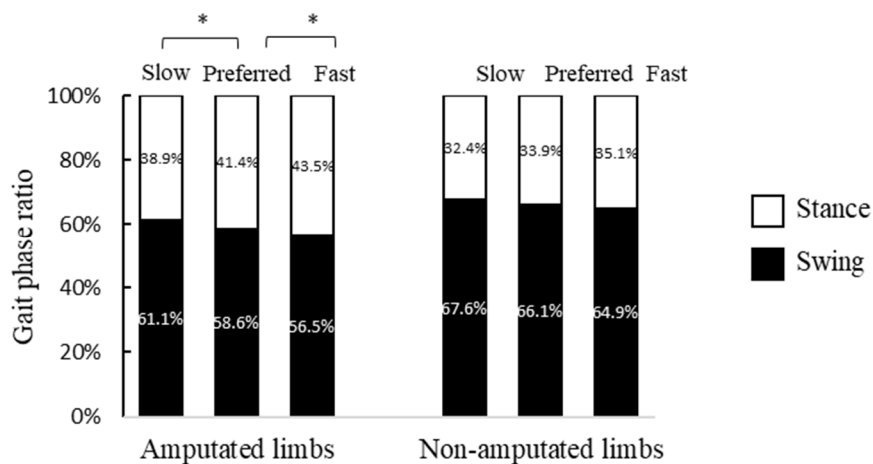
Fig. 6-1 Duration proportion of gait phase under different walking speeds; * indicates that there is a significant difference ($p < 0.05$)

The walking speed remarkably affected the gait phase proportion of amputated limbs, but exerted no significant effect on non-amputated limbs, as shown in Fig. 6-1. Compared with the preferred speed, the proportion of the prosthetic stance phase increased to 61.1% when the amputee walked slowly ($p = 0.03$), but decreased to 56.5% ($p < 0.01$) with fast speed. However, the proportion of the stance phase in non-amputated limbs was kept at approximately 66% in all three speeds.

Table 6-1 Mean and standard deviation of gait parameters of amputees under three walking speeds

Items	Amputated limbs			Non-amputated limbs		
	Slow	Preferred	Fast	Slow	Preferred	Fast
Gait duration (s)	1.44 ± 0.15 (<0.01)	1.30 ± 0.10	1.16 ± 0.13 (<0.01)	1.42 ± 0.14 (<0.01)	1.29 ± 0.10	1.16 ± 0.11 (<0.01)
Walking speed (m/s)	0.77 ± 0.12 (<0.01)	0.91 ± 0.08	1.09 ± 0.14 (<0.01)	0.78 ± 0.12 (<0.01)	0.92 ± 0.09	1.10 ± 0.13 (<0.01)
Stride length (m)	1.11 ± 0.09 (0.04)	1.19 ± 0.03	1.25 ± 0.07 (0.049)	1.1 ± 0.09 (0.06)	1.18 ± 0.06	1.28 ± 0.09 (0.04)
Hip angle at HS (°)	31 ± 6.1 (0.5)	30.2 ± 5.1	30.7 ± 6.9 (0.73)	32.6 ± 2.4 (0.29)	31.4 ± 3.5	33 ± 4.3 (0.18)
Max hip extension angle (°)	−6.7 ± 6.8 (<0.01)	−8.4 ± 6.3	−8.1 ± 8.4 (0.79)	−7.8 ± 6.7 (0.03)	−10.5 ± 5.3	−10.7 ± 7.4 (0.91)
Max knee flexion angle (°)	49.3 ± 12.6 (0.02)	54.9 ± 9.7	59.1 ± 12.8 (0.28)	55 ± 11.9 (0.09)	58.1 ± 10.9	58.2 ± 11 (0.95)
Max dorsiflexion angle (°)	8.4 ± 4.8 (0.17)	8.9 ± 4.6	9.6 ± 5.3 (0.29)	11.1 ± 6.2 (0.03)	8.4 ± 4.8	9.8 ± 4.5 (0.12)
Max hip flexion moment (Nm/BW)	−0.49 ± 0.18 (0.01)	−0.63 ± 0.2	−0.72 ± 0.26 (0.35)	−0.37 ± 0.23 (0.22)	−0.44 ± 0.28	−0.57 ± 0.34 (0.048)

Max knee extension moment in MSt (Nm/BW)	0.07 ± 0.19 (0.57)	0.10 ± 0.23	0.06 ± 0.22 (0.37)	−0.02 ± 0.26 (0.47)	0.03 ± 0.27	0.13 ± 0.34 (0.11)
Max knee flexion moment in TSt (Nm/BW)	−0.39 ± 0.26 (0.35)	−0.31 ± 0.23	−0.36 ± 0.26 (0.58)	−0.54 ± 0.32 (0.41)	−0.48 ± 0.31	−0.53 ± 0.32 (0.48)
Max plantarflexion moment (Nm/BW)	1.14 ± 0.17 (0.924)	1.13 ± 0.19	1.18 ± 0.19 (0.41)	1.3 ± 0.26 (0.68)	1.32 ± 0.22	1.38 ± 0.23 (0.26)

The numbers in brackets indicate the p value of the paired T-test compared with the preferred speed. BW: Body Weight.

Fig. 6-2 illustrates that speed exerted no significant effect on the hip angle at the beginning of the gait cycle. The hip angles of both lower limbs under the three walking speeds were all amounted to 32° at the beginning of gait (see Table 6-1). Compared with the preferred speed, there was no significant difference in the maximum extension angle of the hip joint in gait in fast walking for both lower limbs. However, the maximum hip joint extension angles of the amputated and non-amputated limbs in the slow speed were significantly reduced by approximately 1.7° and 2.7°, compared with those at the preferred speed. In the amputated limbs, they were −6.7° ± 6.8° vs. −8.4° ± 6.3° (p < 0.01) at the slow and preferred speeds respectively, and in the non-amputated limbs they were −7.8° ± 6.7° vs. −10.5° ± 5.3° (p = 0.03) at the slow and preferred speeds, respectively. Compared with the preferred speed, fast walking had no significant difference in the maximum flexion angle of the knee joint in non-amputated and amputated limbs. However, at slow speed, it was significantly reduced by approximately 5.6°, being of 49.3 ± 12.6° compared with 54.9 ± 9.7° at the preferred speed (p = 0.02).
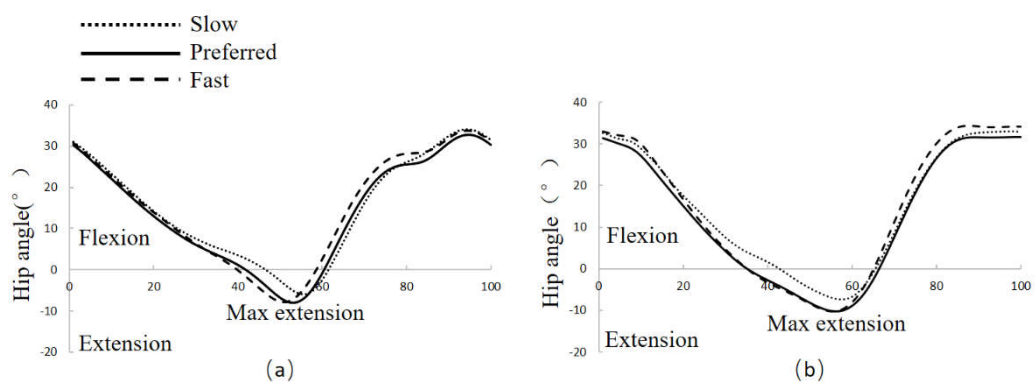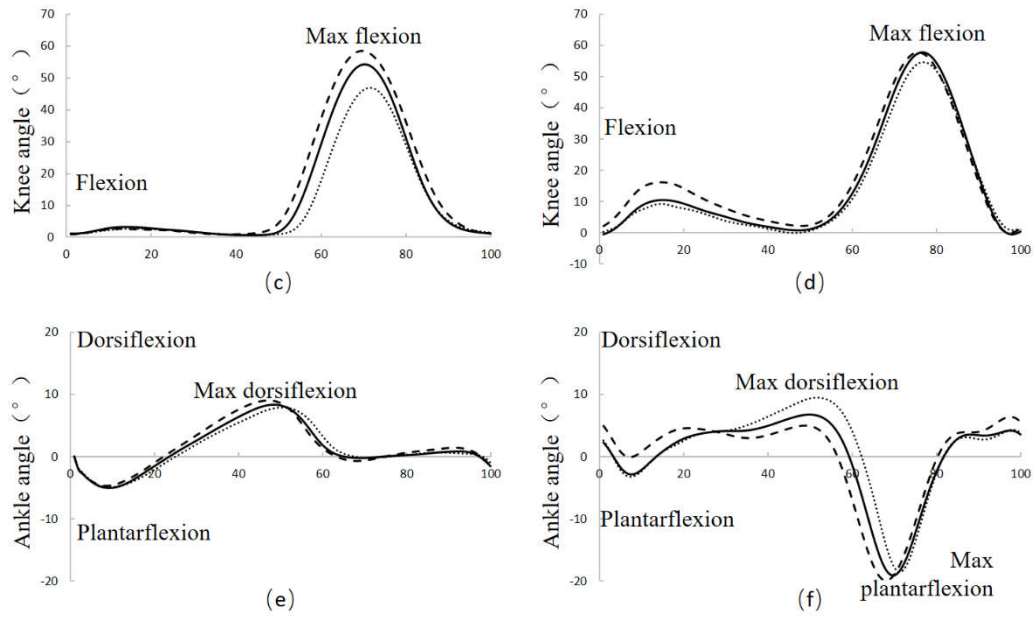
Fig. 6-2 Joint angles in the 100% gait cycle: (a) hip angle of amputated limbs; (b) hip angle of non-amputated limbs; (c) prosthetic knee angle; (d) knee angle of non-amputated limbs; (e) prosthetic ankle angle; (f) ankle angle of non-amputated limbs.

There was no significant difference between the maximum knee flexion angles of non-amputated limbs at the slow and preferred speeds ($p > 0.05$). During fast walking, the maximum plantarflexion angle of the ankle joint of non-amputated limbs was significantly higher than that with the preferred speed, by approximately 2.8°; it was of −0.5° ± 4.6° vs. −3.3° ± 4° ($p = 0.047$), respectively, whereas there was no significant difference in the amputated limbs. There was no significant difference between the maximum ankle flexion angle at the slow and preferred speeds. The effect of speed on the maximum dorsiflexion angle of the ankle joint was only found in the non-amputated limb at slow speed, which was 11.1° ± 6.2° at slow speed and 9.6° ± 5.3° at the preferred speed, respectively ($p = 0.03$).
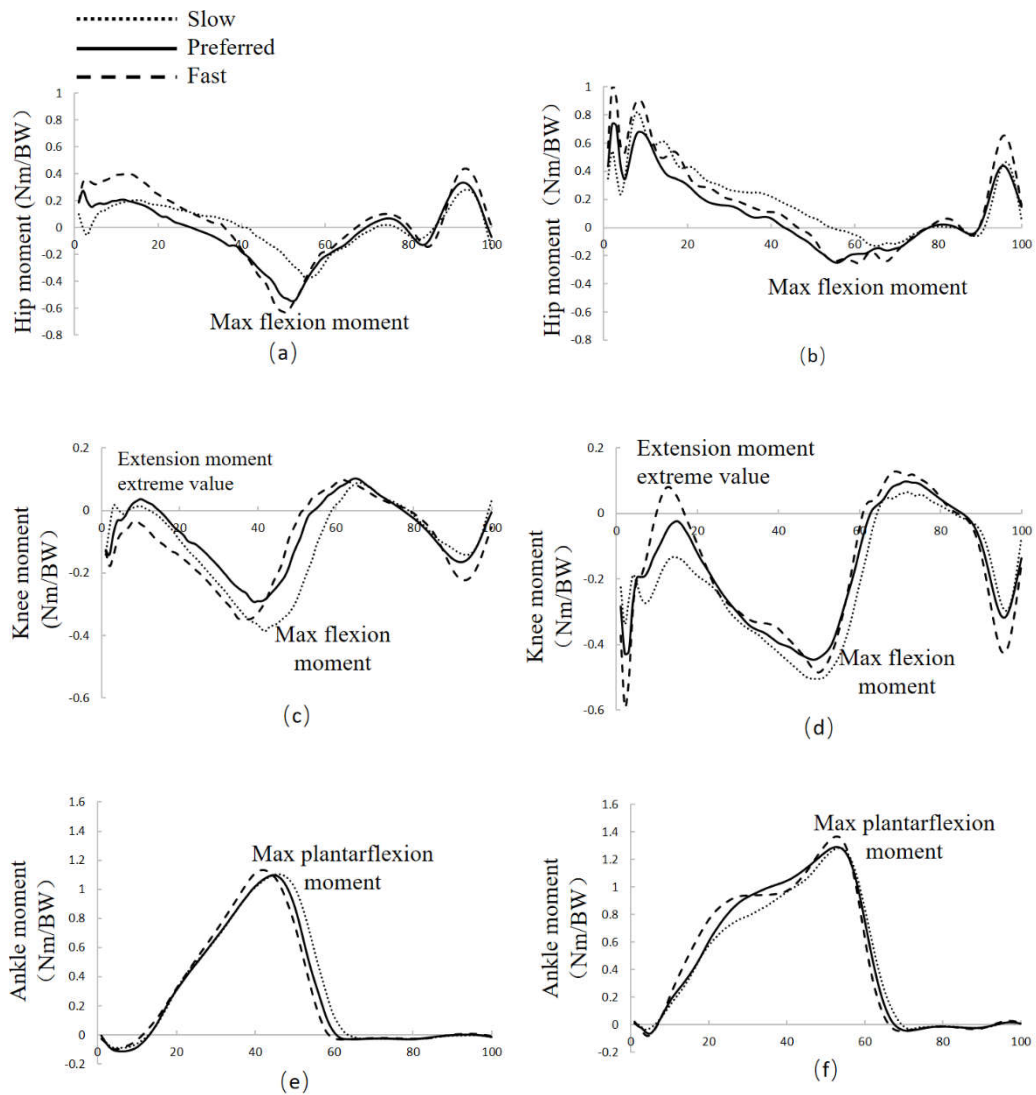
Fig. 6-3 Joint moments in the 100% gait cycle: (a) hip moment of amputated limbs; (b) hip moment of non-amputated limbs; (c) prosthetic knee moment; (d) knee moment of non-amputated limbs; (e) prosthetic ankle moment; (f) ankle moment of non-amputated limbs. BW: Body Weight.

Speed had different effects on the maximum hip joint extension moment of bilateral lower limbs, as shown in Fig. 6-3. The maximum hip flexion moment of the amputated limbs was -0.49 ± 0.18 Nm/BW at slow speed, 0.14 Nm/BW lower than -0.63 ± 0.2 Nm/BW at the preferred speed (p = 0.02); whereas there was no significant difference for the non-amputated limbs at the slow and preferred speeds.

The maximum hip flexion moment of non-amputated limbs at fast speed was −0.57 ± 0.34 Nm/BW, 0.13 Nm/BW higher than −0.44 ± 0.28 Nm/BW at the preferred speed (p = 0.048); whereas there was no significant difference of maximum hip flexion moment for the amputated limbs at the fast and preferred speeds. Speed exerted no significant effect on the maximum knee extension moment in the mid-stance and the maximum knee flexion moment in the terminal stance phase for bilateral lower limbs. There was no significant difference in the maximum plantarflexion moment of the bilateral lower limb ankles at different speeds.

Fig. 6-4 Mean CRP of amputees under different speed gaits: (a) hip–knee of

amputated limbs; (b) hip–knee of non-amputated limbs; (c) knee–ankle of amputated limbs; (d) knee–ankle of non-amputated limbs; (e) hip–ankle of amputated limbs; (f) hip–ankle of non-amputated limbs.

Fig. 6-4 describes the ensemble mean CRP of amputated and non-amputated limbs of amputees under different speed gaits. By comparing the cross-correlation (CC) of inter-joint coordination at slow and fast speeds with that at the preferred speed, it was found that the CC value of the knee–ankle coordination of amputated limbs was smaller than that at the preferred speed in the stance phase, only accounting for 0.70, as presented in Table 6-2. Except for that, the CC values were all greater than 0.85 in inter-joint coordination of the joint groups when walking at fast and slow speeds compared with the preferred speed.

Table 6-2 CC of the CRP when amputees walked fast and slow to compared with preferred speed, respectively.

Joint groups	Gait phase	Slow–Preferred		Fast–Preferred	
		Amputated limbs	Non-amputated limbs	Amputated limbs	Non-amputated limbs
Hip–knee	Stance	0.96	0.88	0.97	0.92
	Swing	0.94	0.97	0.97	0.99
Knee–ankle	Stance	0.94	0.70	0.95	0.85
	Swing	0.87	0.96	0.87	0.97
Hip–ankle	Stance	0.96	0.90	0.94	0.90
	Swing	0.89	0.93	0.90	0.97

Table 6-3 RMS difference of inter-joint coordination between different walking speeds

Joint groups	Gait phase	Slow-Preferred		Fast-Preferred	
		Amputated limbs	Non-amputated limbs	Amputated limbs	Non-amputated limbs
Hip–knee	Stance	2.9 ± 3.1 (0.047)	2.9 ± 2 (0.01)	2.2 ± 5.7 (0.44)	-2.1 ± 5 (0.03)
	Swing	4.4 ± 1.8 (<0.01)	1.5 ± 6.1 (0.55)	-5.4 ± 10.3 (0.09)	1.5 ± 11.8 (0.12)
Knee–ankle	Stance	4.8 ± 4.4 (0.03)	5.8 ± 7.2 (0.08)	3.7 ± 7.9 (0.27)	-1.5 ± 13.5 (0.78)
	Swing	6.6 ± 7.4 (0.06)	3.7 ± 6.3 (0.18)	-10 ± 16.7 (0.22)	0.1 ± 18.8 (0.13)
Hip–ankle	Stance	5.5 ± 8.5 (0.14)	0.9 ± 2.7 (0.41)	-1.3 ± 13.3 (0.11)	-4.8 ± 8.8 (0.64)
	Swing	8.4 ± 11.1 (0.09)	-3.1 ± 6.9 (0.28)	-0.8 ± 16.5 (0.6)	-5.6 ± 14.1 (0.69)

The numbers in brackets indicate the p-value of Paired T-test.

During slow walking, the RMS of the hip–knee CRP on the prosthetic side decreased significantly compared with that of the preferred speed walking in both the stance and swing phases, by approximately 6.8% (p = 0.047) and 6.6% (p < 0.01), respectively. The knee–ankle CRP of the prosthetic side also showed a decrease of 11.8% (p = 0.03), as listed in Table 6-3. When walking slowly, the RMS of the knee–ankle CRP of amputated limbs in the swing phase was approximately 11.2% smaller than that at the preferred speed, and the p-value was 0.06. The RMS of the hip–ankle CRP in the stance and swing phases during slow walking was not significantly different from that at the preferred speed. When walking slowly, the RMS of the hip–knee CRP in non-amputated limbs in the stance phase was reduced

by approximately 6.4% (p = 0.01) compared with that when walking at the preferred speed. No significant RMS difference was found in the hip–knee in the stance phase, and the knee–ankle and hip–ankle in both the stance and swing phases when walking at the slow and preferred speeds. No significant RMS difference was found in non-amputated limbs between fast and preferred speed walking, except for a 4.8% decrease of the hip–knee in the stance phase.

Walking speed had a significant effect on the degree of inter-coordination variations of the amputees, as shown in Fig. 6-5. The DP of hip–knee coordination of amputated limbs at slow speed significantly reduced by 4.3° in the swing phase compared with that at the preferred speed (64.1° ± 10° vs. 68.4° ± 10°, p < 0.01). However, the statistical difference, p-value, of the hip–knee DP of amputated limbs during slow walking was 0.059 (40.3° ± 4° vs. 43.2° ± 2.7°) compared with that at the preferred speed. When walking slowly, the knee–ankle DP of the amputated limbs was significantly different from that when walking at the preferred speed, both in the stance phase (36° ± 6.2° vs. 40.8° ± 6.6°, p = 0.02) and the swing phase (52.8° ± 18.8° vs. 59.7° ± 20.3°，p = 0.05). When walking fast, the hip–knee coordination of amputated limbs was 67.1° ± 9.7° in the swing phase, smaller than the 68.4° ± 10° of that at the preferred speed, p = 0.056.

The degree of variation of hip–knee coordination of non-amputated limbs in the stance phase at slow and fast walking speeds was significantly different from that during walking at the preferred speed. The hip–knee DP during slow walking was 42° ± 3° in the stance phase, lower than 45° ± 4.1° at the preferred speed gait, p = 0.01.
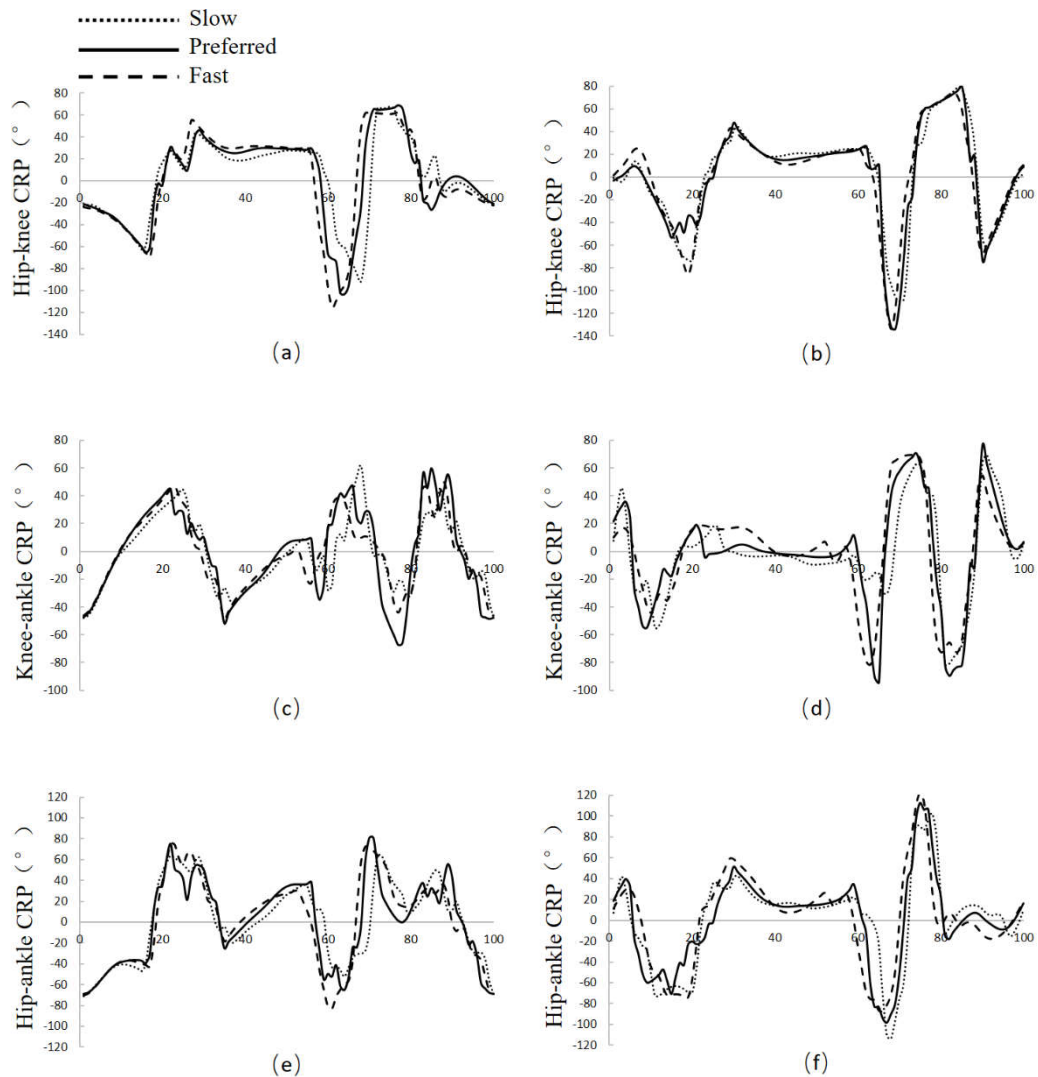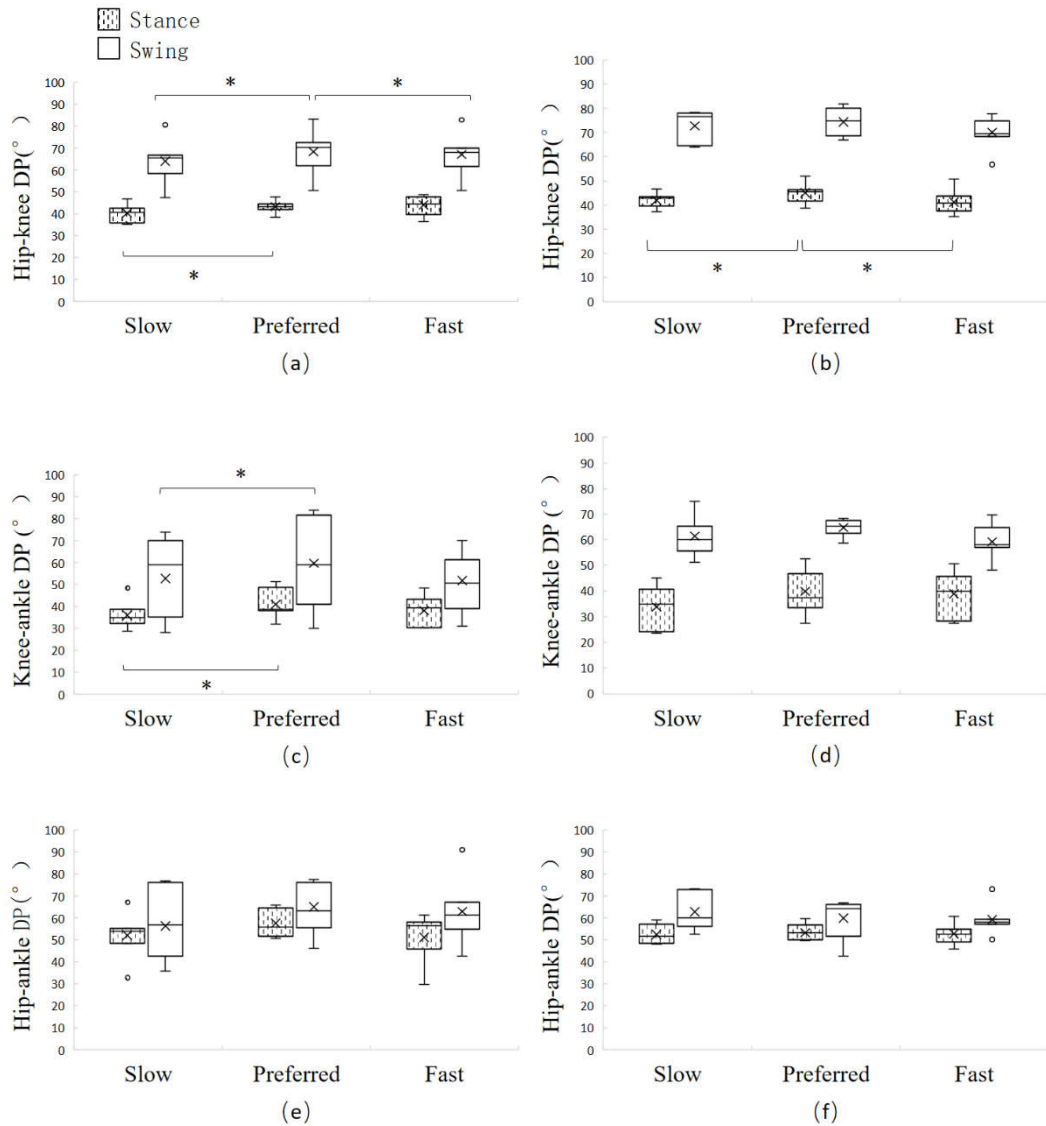
Fig. 6-5 DP between amputees at different walking speeds: (a) hip–knee of amputated limbs; (b) hip–knee of non-amputated limbs; (c) knee–ankle of amputated limbs; (d) knee–ankle of non-amputated limbs; (e) hip–ankle of amputated limbs; (f) hip–ankle of non-amputated limbs; * indicates that there is a significant difference compared with the preferred speed ($p < 0.05$), # means the p-value is between 0.05 and 0.06

The DP of hip–knee coordination of non-amputated limbs was $41.2° \pm 5°$ in the stance phase, significantly lower than the $45° \pm 4.1°$ at the preferred speed, $p = 0.03$.

There was no significant effect of walking speed on the DP of knee–ankle and hip–ankle coordination of non-amputated limbs whatever the phase, stance or swing.

As shown in Fig. 6-6, the DI of the hip–knee in the prosthetic limbs significantly increased when the amputee walked at slow speed compared with that at the preferred speed. The DI at slow speed was $70.7\% \pm 21.8\%$ vs. $60.3\% \pm 20.7\%$ at the preferred speed, $p = 0.05$; The hip–ankle DI increased significantly from $25.3\% \pm 23.3\%$ at slow speed to $19.6\% \pm 17.7\%$ at the preferred speed, $p = 0.05$.

The hip–knee DI of amputated limbs decreased significantly when the amputee walked fast, and it was $50.7\% \pm 18.3\%$ compared with $60.3\% \pm 20.7\%$ at the preferred speed, $p < 0.01$. The DI of the hip–ankle decreased significantly at fast speed, and it was $15.9\% \pm 16.3\%$ vs. $19.6\% \pm 17.7\%$ at the preferred speed, $p = 0.04$. During fast walking, the hip–knee DI of non-amputated limbs was $13.3\% \pm 8.6\%$, significantly lower than $30.4\% \pm 18.2\%$ at the preferred speed, $p = 0.01$. The walking speed made no significant effect on knee–ankle DI whatever the phase, stance or swing.
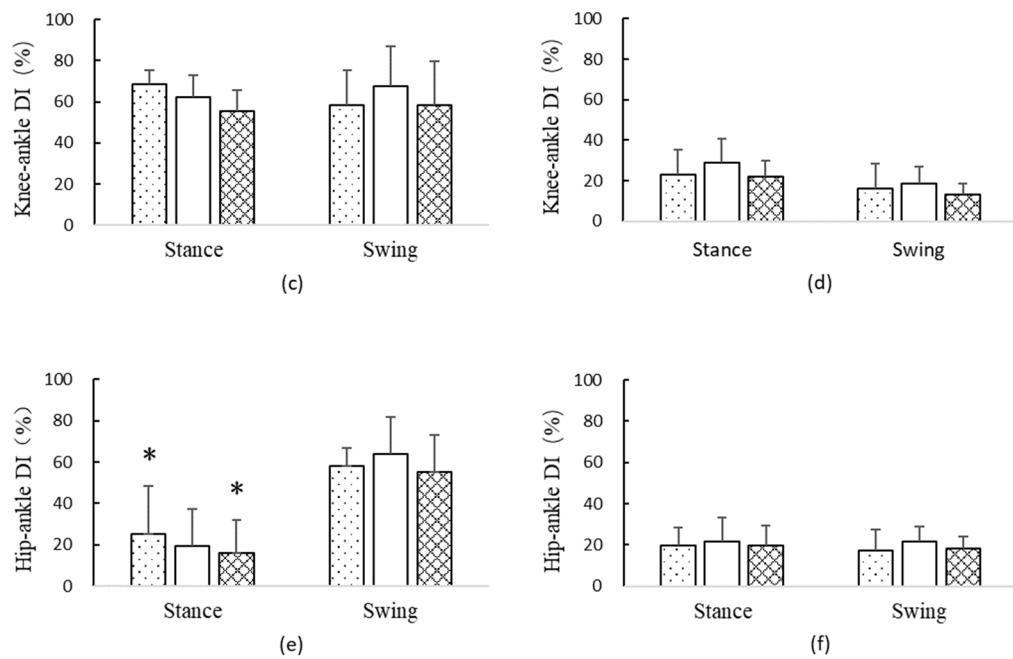
Fig. 6-6 DI of: (a) hip–knee of amputated limbs; (b) hip–knee of non-amputated limbs; (c) knee–ankle of amputated limbs; (d) knee–ankle of non-amputated limbs; (e) hip–ankle of amputated limbs; (f) hip–ankle of non-amputated limbs; * indicates a significant difference compared with the preferred speed ($p < 0.05$)

In this chapter, we compared the inter-joint gait coordination when amputees walked at different walking speeds. The experiment was conducted with unilateral transfemoral amputees at the slow, preferred, and fast walking speeds. By comparing and analyzing the similarity and variation of the inter-joint coordination of amputees at slow and fast speeds with those at the preferred speed, and analyzing them with the DI, the coordination characteristics of amputees at different walking speeds were revealed. The results show that, compared with walking at a comfortable speed (the preferred speed), the lower limb joints of amputees walking at slow and fast speeds are not only different in spatio-temporal parameters, kinematics, and dynamics but also significantly different in gait coordination.

6.2 Conventional gait analysis

Previous studies have found changes in the gait parameters of amputees with unilateral amputation at different walking speeds(Esposito et al., 2015; Schaarschmidt et al., 2012). This study also analyzed the conventional gait analysis parameters for validation of the experiment and results. For example, the results of gait parameters such as gait cycle duration, stride length, joint angle, and joint moment of the amputees' bilateral limbs are consistent with those in previous studies. From the perspective of kinematics, the maximum extension hip angle and maximum knee flexion angle of amputated limbs were reduced by approximately 1.5° and 5.6°, respectively, when the amputee walked at slow speed in this experiment. However, there was no significant difference in the joint angles of amputated limbs between the fast and preferred speed walking. The maximum hip extension angle of the non-amputated limb was also significantly reduced by approximately 2.7° when walking at slow speed compared with that walking at the preferred speed, and there was no significant difference between walking at fast speed and walking at the preferred speed. In addition, the maximum dorsiflexion angle of the ankle of the non-amputated limb when walking at slow speed was significantly increased, by approximately 2.7°, compared with that at the preferred speed. From the change in joint angle, it can be seen that the hip, knee, and ankle joints of both lower limbs have different degrees of change when walking slowly. However, during fast walking, the three joints of bilateral lower limbs showed no significant difference in the above parameters. From the point of view of dynamics, the effect of speed on the joint moment of lower limbs of amputees was different from that on the joint angle. Compared with walking at the preferred speed, only

the hip joint of the prosthetic side had a decrease of approximately 0.14 Nm/BW when walking at slow speed and an increase of approximately 0.13 Nm/BW when walking at fast speed. Except for that, no other significant difference was found, as presented in Table 6-1.

The joint angle and joint moment are indicators of the differences in kinematics and dynamics for a single joint at different walking speeds, but those differences are not sufficient to reveal the overall adjustment of the amputee's control strategies. For different walking speeds, amputees were required to adjust the control strategies to achieve a balanced and coordinated movement. This study has shown in Chapter IV that the gait deviation of the amputee's intact joints was caused by the change of movement control strategies. The coordination analysis can more sensitively find the changes in the coordination relationship between joints. Therefore, how to adjust the amputee's body when dealing with different speed requirements needs to be part of the coordination analysis so as to have a deeper understanding of the known gait deviation. The study in this chapter found that the coordination of the lower limb joints of amputees was different with various walking speeds.

6.3 Effects of walking speeds on the amputees' gait coordination pattern

Different walking speeds enabled the coordination pattern adjustment of hip joint to be the most significant. The hip–knee of the amputated limb and hip–knee of the non-amputated limb coordination pattern (CC and RMS) of the amputee changed greatly at different walking speeds, as presented in Tables 5-1 and 5-2. The hip–knee coordination RMS of the prosthesis during slow walking was significantly

reduced, by approximately 6.8% in the stance phase and 6.6% in the swing phase, compared with that at the preferred walking speed. However, the hip–knee coordination RMS of non-amputated limbs was significantly reduced, by approximately 6.4%, when walking at slow speed compared with that walking at the preferred speed. Winter believed that the hip joint played a more important role in maintaining body balance than the distal joint during walking(Winter, 1995). The research by Pang and Yang also showed that the hip joint movement had a crucial role in the gait time sequence control(Pang and Yang, 2010). Chiu and Chou, through the study of healthy young people, also expressed that the lower limb joints were mainly adjusted through hip–knee coordination when dealing with different walking speeds(Chiu and Chou, 2012). In addition, in Chapter IV in this paper, the important role of the hip joint in motion control has also been confirmed. However, in this chapter, research on different walking speeds found that although transfemoral amputees had partial loss of the musculoskeletal system, the adjustment of gait at different speeds for both their amputated and non-amputated limbs was still controlled and adjusted through coordination of the proximal joints, namely, the hip joints.

Compared with various speeds, the coordination adjustment of amputees when walking slowly was more challenging. Chiu and Chou found that speed left an effect on the coordination of lower limb joints, and the effect was more significant at slow speed(Chiu and Chou, 2012), which coincides with the findings in this work. In this study, the hip–knee and knee–ankle coordination patterns of the amputated limbs during slow walking were significantly different from those during walking at the preferred speed, while the coordination patterns when fast walking were similar to

those during walking at the preferred speed (Table 6-3). Part of the musculoskeletal system was lost on the amputated limb of the transfemoral amputee, and the used passive prosthesis knee joint remained locked in the single support phase and could not be adjusted according to the speed. Similarly, the ankle joint could not be actively adjusted to cope with different speeds owing to the passive working mechanism. Therefore, the active adjustment of motion control to cope with different speeds was mainly carried out through the hip joint, and the role of the hip joint is particularly critical. When walking slowly, the gait duration is longer (Table 6-1), and the proportion of the stance phase of amputated limbs is larger (Fig. 6-1), resulting in longer duration of the prosthetic limb stance phase. The research presented in Chapter IV has found that amputees with a single support phase of the amputated limbs had weak control over movement, and thus, it was necessary for the non-amputated limbs to shorten the swing phase and enter the double support phase as soon as possible to ensure a stable gait. Chiu and Chou's research has identified the typical effect of slow speed on motion control(Chiu and Chou, 2012). They found that not only the hip–knee coordination of the elderly, but also that of young adults, exhibited significant changes when walking at slow speed. They also believed that slow walking was more difficult to control than the preferred and fast walking. However, in our study, the movement adjustment ability of the amputated limb of amputees was lower than that of abled persons; thus, it is more difficult to control gait balance and stability when the duration of the single support phase is prolonged.

The coordination of the prosthetic knee–ankle joint is significantly affected by speed. Especially when walking at slow speed, the knee–ankle coordination RMS

of the prosthesis was significantly reduced, by approximately 11.8%, compared with that of walking at the preferred speed. No significant effect of speed on knee–ankle coordination was found in Chiu and Chou's studies on abled persons(Chiu and Chou, 2012). In this study, the knee–ankle coordination of the non-amputated limb of the amputee was similar when walking at a slow or fast speed compared with that of the preferred speed, but the knee–ankle coordination pattern of amputated limbs changed significantly when walking at slow speed with prostheses, compared with the pattern when walking at the preferred speed. This difference was mainly caused by different mechanical designs of the prosthetic products. The prosthetic knee and ankle joints used by the subjects participating in this experiment were passive prosthetic joints. The difference in coordination pattern reflected in slow walking was not the active adjustment to different speeds, but the different working mechanisms under different conditions. The designs of the prosthetic knee and ankle joint were usually relatively independent, and they lack the ability to coordinate with each other in gait to achieve better coordination. This is a defect of the traditional passive transfemoral prosthesis. Given that walking at different speeds is a daily activity, whether the knee–ankle coordination can adapt to different speeds should also be one of the evaluation criteria for prostheses. Consistent with the previous discussion, the knee–ankle coordination of amputated limbs in slow walking was significantly different from that in the preferred walking, but there was no significant difference in knee–ankle coordination between the fast and preferred walking. For the same reason, to achieve coordination of the amputated limbs became more difficult after the duration of the single support phase increased. Thus, special attention should be paid to the knee–ankle coordination with slow-speed walking. In this study, the p-value in statistics of knee–ankle coordination between

the slow and preferred speeds at the swing phase was 0.06, which is higher than the statistical difference threshold of 0.05. However, considering that, as described in Chapter IV, it was found that the knee–ankle coordination of prostheses was significantly different from that of abled persons in the swing phase, and Nolan et al. believe that the asymmetry of the swing phase is too high to affect the increase in step frequency, thus limiting the improvement in walking speed(Nolan et al., 2003), we believe that the knee–ankle coordination of the amputated limb is also worthy of special attention in the swing phase when walking slowly. The gait deviations of amputees are mainly caused by the large difference between the biomechanical function compensated by artificial limbs and that of abled persons. Therefore, an important standard should be that the knee–ankle coordination of transfemoral prosthesis be reasonably compensated. The rehabilitation of amputees and the verification of prosthesis design should take the knee–ankle coordination as one of the criteria.

It can be seen from the coordination pattern that although the non-amputated limb of the amputee also depends on the proximal joint for adjustment when dealing with different speeds, the adjustment only occurs in the stance phase (the hip–knee coordination RMS of the non-amputated limb increases by 6.4% at slow speed and decreases by 4.8% at fast speed, compared with that at the preferred walking speed), while in the swing phase there is no significant difference in coordination. Chiu and Chou, through the study of abled persons, considered that the adjustment of joint coordination in the stance phase was more important than the adjustment of the swing phase when dealing with speed changes(Chiu and Chou, 2012). It can be observed that the non-amputated limbs of amputees with unilateral transfemoral

amputation have a similar joint coordination adjustment mechanism to that of abled persons when dealing with different walking speeds.

6.4 Changes in the degree of variation in the coordination among amputees under various speeds

Judging from the DP value, the preferred walking speed that amputees feel comfortable with is an embodiment of high joint control flexibility, and it is the "optimal" speed. Compared with walking at the preferred speed, the hip–knee and knee–ankle DP of amputated limbs and the hip–knee DP of non-amputated limbs showed significant reduction when the amputee walked at slow and fast speeds, as shown in Fig. 6-5. In particular, the p-value of hip–knee coordination of amputated limbs at the stance phase between the slow and preferred walking was 0.059, and of hip–knee coordination of non-amputated limbs at the swing phase between the fast and preferred speeds was 0.056. Both cases exceeded the significance determination threshold of 0.05. However, considering that the researchers have confirmed that speed played a significant effect on hip–knee coordination variability in gait(Chiu and Chou, 2012), and that we have found that speed has a significant effect on the coordination pattern of amputees, we believe that the variation in degree of coordination in these two states deserves the same attention with significant differences. In this study, the DP was reduced when there was a significant difference compared with that when walking at the preferred speed. That is, when walking at a speed deemed comfortable by oneself, there is a large difference in joint coordination between amputees. However, when walking at a slow or fast speed, the difference in joint coordination between amputees was relatively small. The variation degree in joint coordination was a manifestation of

joint control flexibility(Kemoun et al., 2002; Scholz, 1990). Therefore, it indicated at different speeds that amputees possessed a more flexible joint control at the preferred walking speed. The flexibility of adjustment of joint coordination experienced a reduction when walking at slow and fast speeds.

Genin et al. explained the "U" relationship between walking speed and metabolic consumption of amputees with transfemoral amputation, and confirmed the existence of an "optimal" walking speed(Genin et al., 2008). That is, when walking at the "optimal" speed, the metabolic consumption is the lowest, and decreasing or increasing the walking speed will bring about an increase in metabolic consumption. The increase in metabolic consumption indicates that exercise is more difficult, and thus, less flexible. Therefore, the conclusion of Genin et al. is consistent with the finding in this study. This means that the amputee has an "optimal speed" at which the amputee can more easily control the gait movement. Walking at this speed is the easiest and most comfortable.

Additionally, both the present study and the research from Genin et al. found that the optimal speed of amputees with transfemoral amputation was remarkably reduced in comparison with that of abled persons. In this study, the preferred speed at which amputees feel comfortable is approximately 0.45 m/s lower than that of the control group, and there is even a greater difference with abled persons with regard to movement ability.

6.5 Effects of velocity on the decomposition movement

The DI revealed that the amputee's motion control was easier when walking fast, but more difficult when walking slowly. DI refers to the movement of one joint

while the other remains fixed(Bastian et al., 1996). In this study, the hip–knee and hip–ankle DI index in the stance phase of the amputee's amputated limbs decreased with the increase in speed, while in the swing phase it has no significant difference (Fig. 6-6). With the increase in walking speed, the duration of the amputee's gait cycle (Table 6-1) and the proportion of the stance phase decreased (Fig. 6-1), while the stride length increased. The amputated limb needs to complete the stance phase in a shorter time owing to the athletic ability weakness in the single support phase; hence, the movement of the hip, knee, and ankle joints on the prosthetic side accelerated, and thus, the duration decreased. The increase in joint movement speed significantly reduced the decomposition of prostheses. On the contrary, the decomposition movement was significantly increased when walking slowly. In Chapter IV we have pointed out that the increase in the decomposition movement was considered as an expression of a compensatory mechanism. The musculoskeletal system of the lower limbs is characterized by a high degree of freedom, and the adjustment of joints and body parts in gait is extremely complicated. The decomposition movement can effectively reduce the degree of freedom in the control system(Bastian et al., 1996; Earhart and Bastian, 2001), making the coordination adjustment simpler and more effective. The knee and ankle joints of a transfemoral prosthesis belong to prosthetic joints, which are driven by the hip joint in gait to adapt to different walking speeds. The change in the movement speed of the prosthetic knee and ankle joint is a passive change. The DI index decreased with the increase in speed, which proved that it was easier to control the motion while walking fast, whereas the control difficulty increased while walking slowly.

The DI index also evidenced that the adjustment of the amputated limb control strategies under different speeds was mainly carried out through the hip joint. With the increase in speed, the DI index of the prosthetic hip–knee and hip–ankle presented significant decreasing trends. In addition, the motion adjustment mainly occurred in the stance phase, while the adjustment in the swing phase was relatively small. In this study, the DI index of the prosthetic hip–knee and hip–ankle decreased remarkably in the stance phase with the increase in speed, but there was no significant difference in the swing phase. This point was also consistent with the Chiu and Chou's point of view discussed above, which holds that the joint coordination in the stance phase plays a more important role when the speed changes.

It can be observed from the proportion of the swing phase (Fig. 6-1) that, while the duration of the gait cycle decreases with the increase in walking speed (Table 6-1), the proportion of the swing phase of amputated limbs failed to remain the same, but increased with a higher speed. Actually, there is a high asymmetry between the prosthesis and the non-amputated limb in amputees with unilateral transfemoral amputation(Schaarschmidt et al., 2012). Nolan et al. stated that to increase the walking speed of amputees, the asymmetry of the gait of prostheses and non-amputated limbs in the swing phase will be reduced; otherwise, the increase in step frequency will be limited, thus making it difficult to increase the walking speed(Nolan et al., 2003). However, the results of this study revealed that it is precisely because of the limitation of the amputees' movement ability that it is difficult for the amputee's prosthetic limb to significantly shorten the duration of the swing phase. Therefore, the proportion of the swing phase increases with the

increase in speed, resulting in the high asymmetry between the two lower limbs, thus limiting the increase in step frequency and making it difficult for amputees to reach a higher walking speed. The non-amputated limb has a sound musculoskeletal system and abled nerve control ability. When walking at the preferred speed, the non-amputated limbs DI was significantly different from that of abled persons in only hip-ankle (refer to the finding in Chapter IV). Whereas, the hip–ankle DI of non-amputated limbs did not vary significantly at different speeds. In addition, DI difference of non-amputated limbs at different speeds was only found in hip-knee, at which the amputees walking fast. Therefore, speed has a small effect on the decomposition movement of the non-amputated limbs. Beyond the change in hip–knee coordination pattern, the change in hip–knee DI during fast walking also indicates the regulating function of the hip joint for coping with different speeds.

6.6 Summary

This chapter aims to study the gait coordination of transfemoral amputees at different walking speeds. A gait analysis was conducted with unilateral transfemoral amputees at the preferred, slow, and fast speeds. The results of the coordination study revealed that the change in the motion control mechanism of amputees at various speeds, and it demonstrated the sensitivity of gait coordination analysis in detecting gait deviation. There were significant differences in the inter-joint coordination with different velocities. The main findings mentioned in this chapter are as follows:

The hip joints of both lower limbs play a crucial role in motion control to cope with different speeds. The coordination adjustment of non-amputated limbs for different

speed requirements mainly occurs in the stance phase. Among different speeds, the challenge of motion control and coordination adjustment is more difficult for amputees who walk at a slower speed. Speed has a significant effect on the prosthetic knee–ankle coordination. Currently, the knee–ankle coordination of a transfemoral prosthesis applied in the clinic has a weak adaptability to different walking speeds. Therefore, it is necessary to strengthen the adaptability of the knee–ankle coordination to different walking speeds in future designs. The inter-joint coordination of both non-amputated and amputated limbs is affected by speed. However, non-amputated limbs show a regulation mechanism similar to abled persons. The preferred speed at which amputees feel comfortable is the "optimal" speed at which the motion control is more flexible.

It is inevitable for amputees to walk at different speeds, and the ability of motion control of amputees is insufficient. The research discussed in this chapter shows that the hip joints of both lower limbs is of great significance for coping with different speeds. Targeted training of the hip joints would improve the amputee's control over walking speed. The prosthesis design also needs to be improved according to the different characteristics of the amputees' gait coordination at different speeds, such as the different requirements of knee–ankle coordination, so as to ensure better gait coordination.

CHAPTER VII MUSCLE COORDINATION OF THE AMPUTEES' NON-AMPUTATED LIMBS

This chapter aims to present the characteristics of muscle coordination of the non-amputated limbs of unilateral transfemoral amputees. Muscle coordination was defined as the distribution of muscle activity or muscle force among muscles in order to produce a specified combination of joint moments (Prilutsky, 2000). The importance of muscles lies in their ability to generate and regulate joint moments required to drive particular movements (Levine et al., 2012). Joint moment has frequently been used to evaluate gait performance but is insufficient for this purpose if muscle forces are not considered, because similar joint moments can be induced with various levels of muscle co-contraction. For example, the joint moment produced by flexors can be offset by extensors. Inoue et al. (2011) observed that this agonist–antagonist activity can provide better stability by increasing joint stiffness. Thus, the understanding of non-amputated limb muscle coordination could help in the understanding of the motor control strategy of non-amputated limbs for ensuring the gait coordination.

In this study, the amputees wore the passive transfemoral prostheses that they use in their daily lives and walked at their preferred speed. The 24 muscles in the non-amputated limb were extracted to calculate the muscle forces in gait through an inverse dynamics analysis and compared with those of abled persons. The inverse dynamic analysis discussed in this chapter is based on the gait experiment conducted in Chapter IV. The calculation process of inverse dynamics was explained in Section 3.2, Chapter III. In this chapter, the muscle coordination will

be discussed in conjunction with joint moment.

According to the definition of the joint muscle from Perry(Perry et al., 2010), 24 groups of muscle bundles were extracted from the lower limb, including:

1) Five groups of hip extensors: biceps femoris caput longum (BFL), semitendinosus (ST), semimembranosus (SM), adductor magnus (AM), and gluteus maximus (GMa);

2) Five groups of hip flexors: adductor longus (ADL), rectus femoris (RF), Gracilis (GRa), sartorius (SAr), and iliacus (ILi);

3) Four groups of knee extensors: vastus lateralis (VL), vastus medialis (VM), vastus intermedius (VI), and rectus femoris (RF);

4) Three groups of knee flexors: biceps femoris caput breve (BFB), popliteus (POp), and gastrocnemius (GAs);

5) Six groups of ankle plantar flexors: soleus (SOl), gastrocnemius (GAs), tibialis posterior (TP), flexor digitorum longus (FDL), flexor hallucis longus (FHL), and peroneus (Per);

6) Three groups of ankle dorsiflexors: tibialis anterior (TA), extensor hallucis (EH), and extensor digitorum (ED).

The rectus femoris is defined as both hip flexor and knee flexor, and the gastrocnemius is defined as both knee flexor and ankle plantar flexor.

In this chapter, two characteristic instants were selected for the discussion of the hip, knee, and ankle moment of the lower limbs:

1) H1: maximum extension moment of the hip joint in the loading response

phase;

2) H2: maximum extension moment of the hip joint in the initial swing phase;

3) K1: maximum extension moment of the knee joint in the mid-stance phase;

4) K2: maximum flexion moment of the knee joint in the terminal stance phase;

5) A1: maximum dorsiflexion moment of the ankle joint in the loading response phase;

6) K2: maximum plantar flexion moment of the ankle joint in the terminal stance phase.

Independent Sample T-test was used to compare the differences in the parameters (including the walking speed, joint angle, joint angular velocity, joint moment, and distance from joint moment to rotation center) between two groups; A *pre-hoc* test, as assessed by Shapiro-Wilk's test, demonstrated that the muscle forces were not normally distributed. Therefore, a Mann-Whitney U test was carried out to compare the muscle forces of two groups. The significance level was set at $\alpha=0.05$.

7.1 Results of inverse dynamics analysis

By comparing the data of the control group with the research results from other scholars, the muscle force of the lower limbs extracted by an inverse dynamics analysis in this study was in good agreement with the measurement results of the EMG technology in other studies, regardless of peak value or pattern(Bogey et al., 2005; Heintz and Gutierrez-Farewik, 2007; Lim et al., 2017). Considering that how to obtain an accurate muscle force through the EMG technology is still under

discussion among scholars(Wentink et al., 2013; Wong et al., 2015), this study mainly carried out a qualitative analysis and discussion of the comparison between the amputees' non-amputated limbs muscle force and that of abled persons, so as to present the characteristics of the muscle coordination of unilateral transfemoral amputees.

Table 7-1 Spatio-temporal parameters of gaits for non-amputated limbs of amputees and the control group

Parameters	Control group	Non-amputated limbs of amputees	P value
Walking speed (m/s)	1.37 ± 0.13	0.92 ± 0.08	<0.01
Stride length (m)	1.37 ± 0.13	1.21 ± 0.06	<0.01
Gait duration (s)	1.11 ± 0.04	1.29 ± 0.08	<0.01
Stance phase (%)	60.5 ± 1.7	66.1 ± 2.6	<0.01

Results indicated that the walking speed of amputees was slower. The walking speed and stride length of the non-amputated limb gait were significantly reduced by approximately 15.3% and 11.7%, respectively, compared with the control group, as shown in Table 7-1. Meanwhile, the duration of the gait cycle increased by 16.2% compared with that of the control group. The stance phase of non-amputated limbs of amputees was significantly longer, accounting for $66.1\% \pm 2.4\%$ of the gait cycle. The proportion of the abled persons' stance phase amounted to $60.5\% \pm 1.7\%$, which was close to the golden ratio proposed by Iosa(Iosa et al., 2013).

There was no significant difference in hip moment between non-amputated limbs of amputees and that of abled persons at H1 instant (0.81 ± 0.29 Nm/BW vs. 0.87 ± 0.22 Nm/BW, $p > 0.05$); However, at H2 instant, the hip flexion moment of amputees' non-amputated limbs was reduced by approximately 58% compared with

that of abled persons (-0.44 ± 0.27 Nm/BW vs. -1.05 ± 0.45 Nm/BW, p < 0.01), as shown in Fig. 7-1. A further investigation showed that the angular velocity of hip joint of amputees is different from that of the control group at H2 instant, accounting for -12.2 ± 24.7 and 56.3 ± 49.9 (p < 0.01), respectively.

(a)

(b)

Fig. 7-1 Ensemble mean hip angle and hip moment within 100% gait cycle: (a) joint angle; (b) joint moment. ☆ indicates that there is a significant difference between the non-amputated limbs of amputees and the control group. H1 is for the maximum extension moment of the hip joint in the loading response; H2 is for the maximum extension moment of the hip joint in the pre-swing. The shaded areas indicate ± 1 standard deviation from the mean

Compared with the lower limbs of abled persons, the intact knee joint of amputees showed smaller extension moment (0.04 ± 0.32 Nm/BW vs. 0.49 ± 0.3 Nm/BW, p

< 0.01) at K1, but higher flexion moment (-0.44 ± 0.32 Nm/BW vs. -0.09 ± 0.26 Nm/BW, p = 0.03) at K2, as shown in Fig. 7-2. The non-amputated limb of amputees exhibited lower knee joint extension moment or higher flexion moment in majority of stance phases (from the loading response to the end of the stance phase). K2 occurred in the terminal stance, and the distance from the GRF vector to the center of knee joint rotation was 46.69 ± 29.59 mm in the gait of amputees, which was significantly greater than 9.78 ± 27.21 mm in abled persons (p = 0.03).
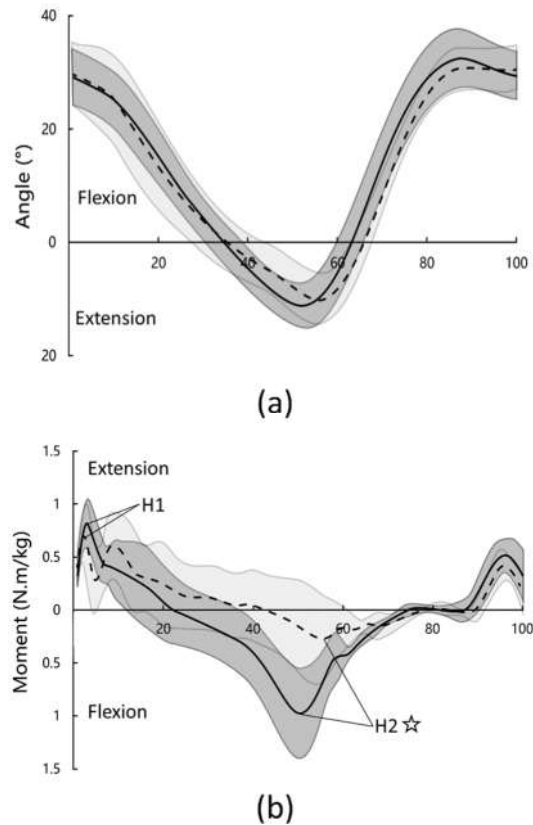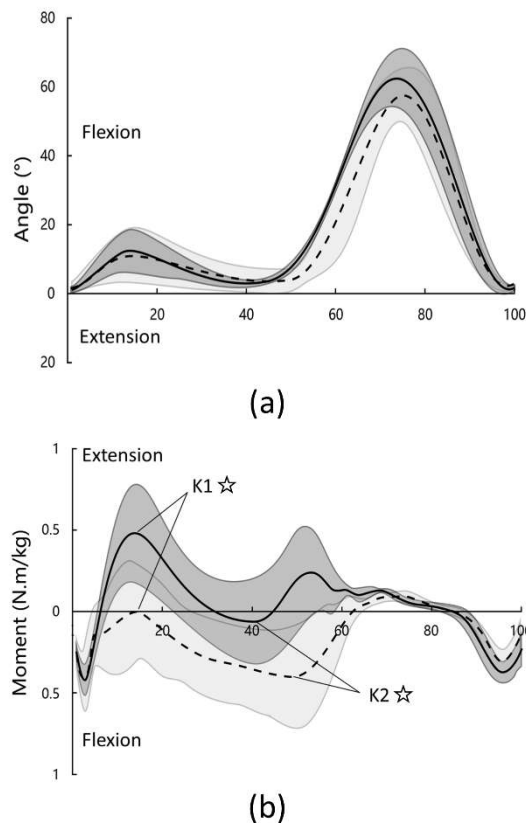

Fig. 7-2 Ensemble mean knee angle and knee moment within 100% gait cycle: (a) joint angle; (b) joint moment. ☆ indicates that there is a significant difference between the non-amputated limbs of amputees and the control group. K1 is for the maximum extension moment of the knee joint in the mid-stance; K2 for the maximum flexion moment of the knee joint in the terminal stance. The shaded areas indicate ± 1 standard deviation from the mean

Subjects in the two groups exhibited no significant difference at both A1 and A2 with regard to their ankle moment. As shown in Fig. 7-3, the ankle moment of non-amputated limbs of amputees and lower limbs of abled persons was A1 (1.27 ± 0.24 Nm/BW vs. 1.36 ± 0.21 Nm/BW, $p > 0.05$) and A2 (-0.11 ± 0.16 Nm/BW vs. -0.14 ± 0.05 Nm/BW, $p > 0.05$).

(a)

(b)
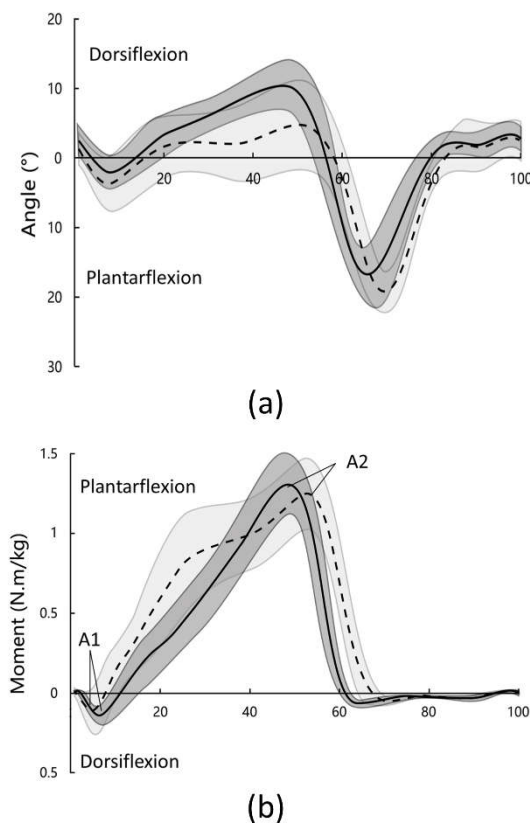
Fig. 7-3 Ensemble mean ankle angle and ankle moment within 100% gait cycle: (a) joint angle; (b) joint moment. A1 is for the maximum dorsiflexion moment in the loading response; A2 for the maximum plantarflexion moment of the ankle joint in the terminal stance. The shaded areas indicate ± 1 standard deviation from the mean

The non-amputated limbs muscle force of amputees and that of the control group were calculated by an inverse dynamic simulation. The average muscle force of hip extensors (Fig. 7-4), hip flexors (Fig. 7-5), knee extensors (Fig. 7-6), knee flexors

(Fig. 7-7), ankle plantar flexors (Fig. 7-8), and ankle dorsiflexors (Fig. 7-9) are shown in the corresponding figures.
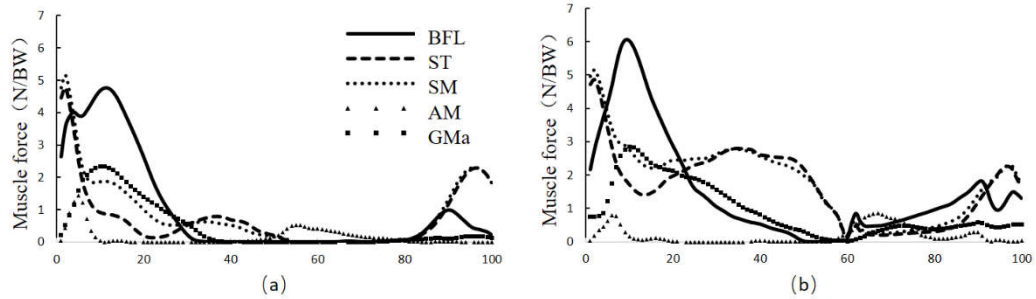
Fig. 7-4 Ensemble mean hip extensors' muscle forces within 100% gait cycle: (a) non-amputated limb; (b) control group
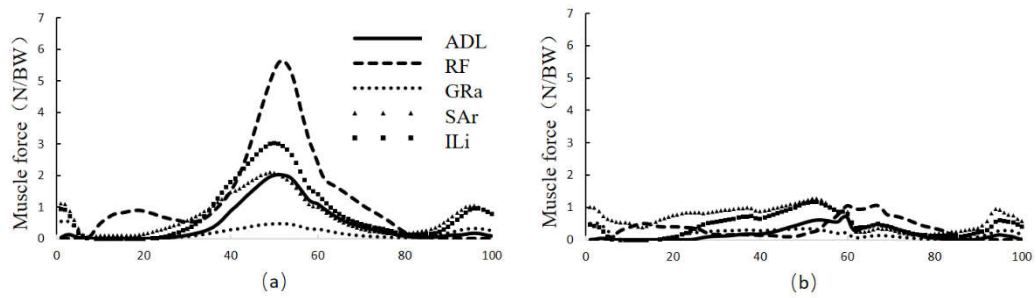
Fig. 7-5 Ensemble mean hip flexors' muscle forces within 100% gait cycle: (a) non-amputated limb; (b) control group
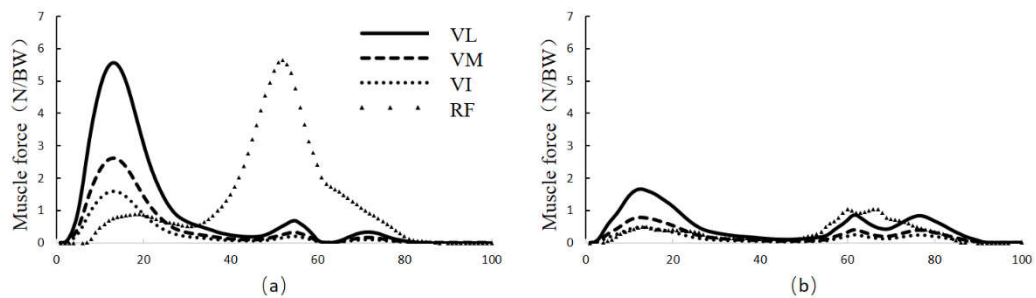
Fig. 7-6 Ensemble mean knee extensors' muscle forces within 100% gait cycle: (a) non-amputated limb; (b) control group
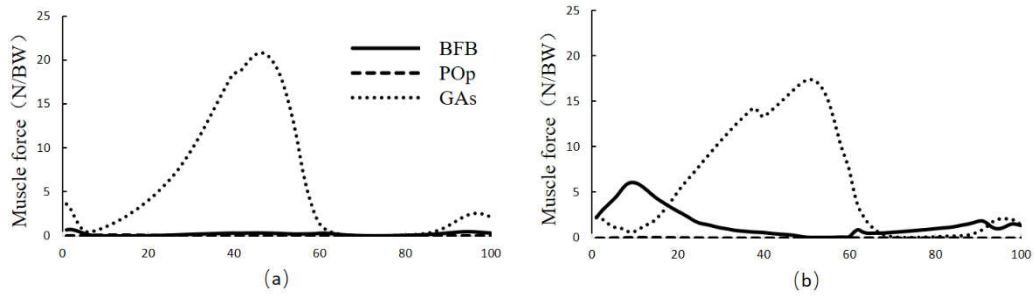
Fig. 7-7 Ensemble mean knee flexors' muscle forces within 100% gait cycle: (a) non-amputated limb; (b) control group
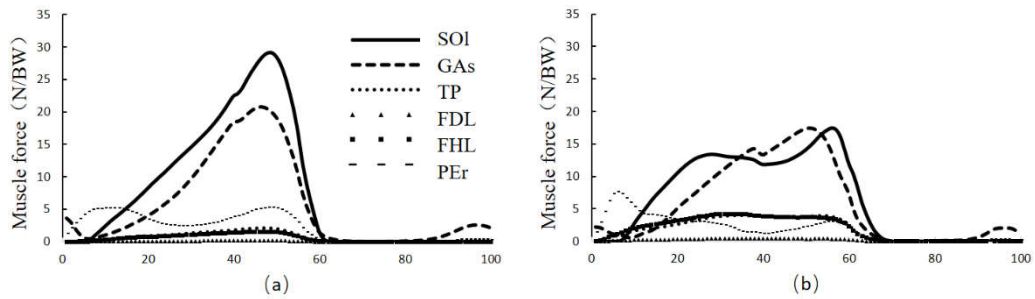
Fig. 7-8 Ensemble mean ankle plantarflexors' muscle forces within 100% gait cycle: (a) non-amputated limb; (b) control group
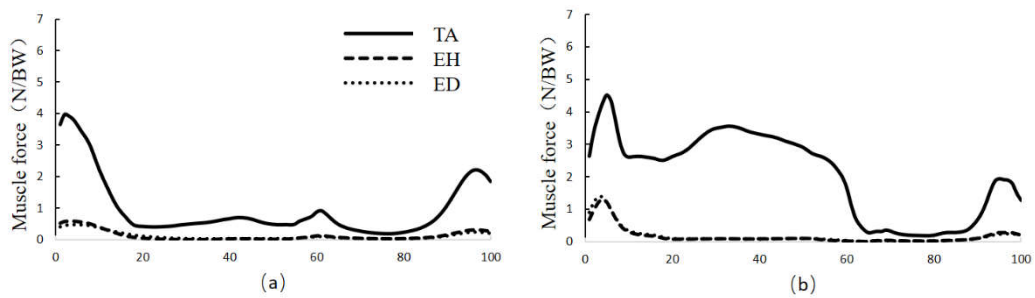
Fig. 7-9 Ensemble mean ankle dorsiflexors' muscle forces within 100% gait cycle: (a) non-amputated limb; (b) control group

The forces of the corresponding muscles of subjects at each characteristic instant were estimated to obtain the muscle coordination as shown in Fig. 7-10, 7-11, and 7-12.

Fig. 7-10 Hip muscle forces at: (a) H1 instant; (b) H2 instant. ☆ indicates that there is a significant difference between the non-amputated limb of amputees and the control group.

All hip flexors and extensors of non-amputated limbs of amputees showed no significant difference with those of abled persons at H1 (p > 0.05), as shown in Fig. 7-10 a. Except the GRa, all hip flexors of non-amputated limbs of amputees displayed lower muscle forces than those of abled persons at H2, as shown in Fig. 7-10 b. Among them, compared with abled persons, the ADL of amputees was 58% smaller (p = 0.03), the RF 81% smaller (p < 0.01), the SAr 49% smaller (p = 0.02), and the Ili 63% smaller (p = 0.01). In addition, there was no significant difference in all the hip extensors of non-amputated limbs of amputees and abled persons.
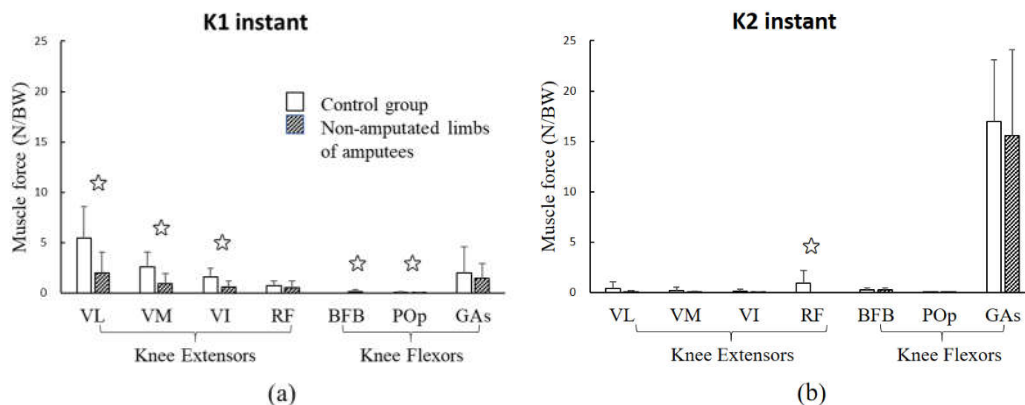
Fig. 7-11 Knee muscle forces at: (a) K1 instant; (b) K2 instant. ☆ indicates that there is significant difference between the non-amputated limb of amputees and the control group.

For the knee joint muscles of non-amputated limbs, the muscle forces of the quadriceps except the RF were significantly reduced at K1 compared with that of abled persons, as shown in Fig. 7-11 a. Moreover, the VL, VM, and VI were respectively 64% (p = 0.04) smaller. The muscle forces of the non-amputated limbs knee flexors were also different from that of the control group. The muscle force of the POp was only 0.027 ± 0.027 N/BW, reduced by 62% (p = 0.02), but the BFB was larger than that of the control group (p = 0.04). At K1 instant, there was little muscle force for the BFB in the control group, while the force of the non-amputated limbs of amputees was approximately 0.12 ± 0.2 N/BW. At K2 instant, only the RF was significantly different from that of the control group (p < 0.01) in terms of muscle force, as shown in Fig. 7-11 b. In the control group, it approximately amounted to 0.94 ± 1.2 N/BW, while there was little muscle force for amputees.
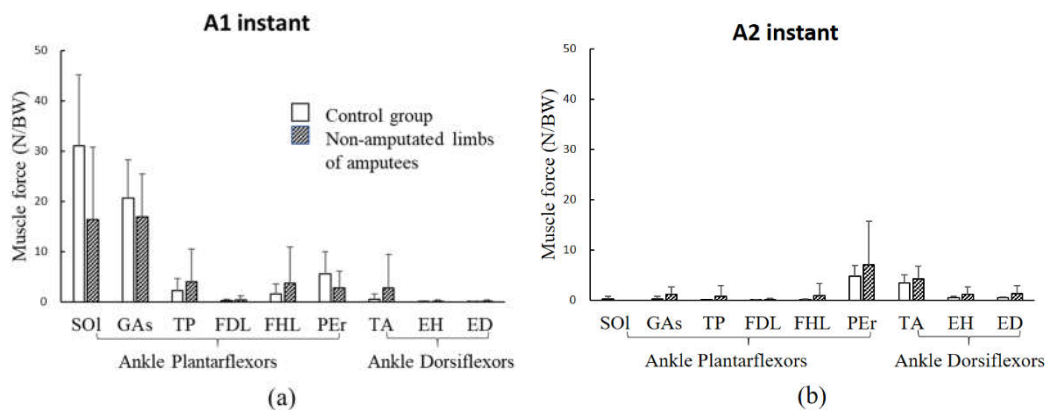
Fig. 7-12 Ankle muscle forces at: (a) A1 instant; (b) A2 instant. ☆ indicates that there is significant difference between the non-amputated limb of amputees and the
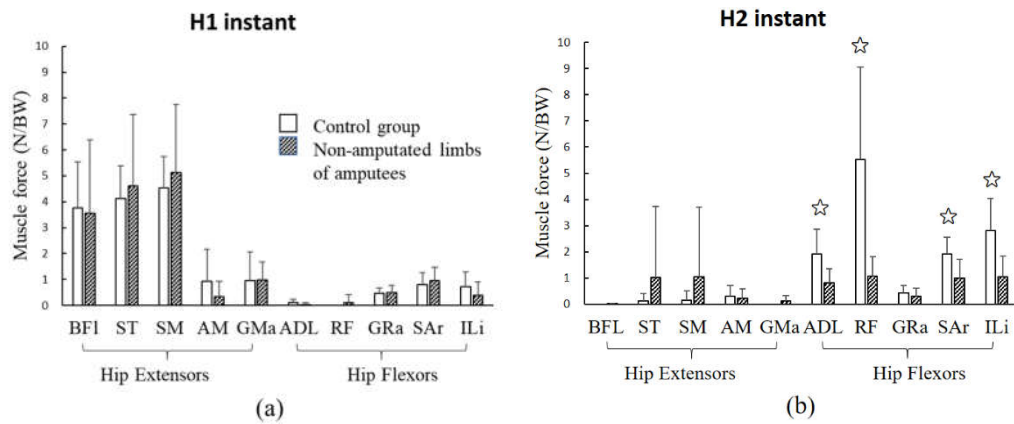
control group.

The amputees' dorsiflexors and plantarflexors presented no significant difference with those of abled persons both at A1 and A2 (p > 0.05), as shown in Fig. 7-12 a and Fig. 7-12 b.

The maximum forces of each muscle were also considered. Fig. 7-13 describes the maximum muscle forces of the non-amputated limbs of amputees and those of the control group, respectively.

There was no significant difference in the maximum muscle force of 5 hip extensors, 3 knee extensors, 6 ankle plantar flexors, and 3 ankle dorsiflexors between the non-amputated limbs of amputees and the control group (p > 0.05), as shown in Fig. 7-13 a, Fig. 7-13 c, Fig. 7-13 e, and Fig. 7-13 f. However, Fig. 7-13 b and Fig. 7-13 c demonstrate that the maximum muscle force of the RF of amputees' non-amputated limbs reached 1.73 ± 0.68 N/BW, 72.9% (p < 0.01) less than that of the control group, and the force of another ankle flexor, the ILi of amputees' non-amputated limbs, was 1.73 ± 1.01 N/BW, 45.4% (p = 0.03) less than that of abled persons.

(a)

(b)

(c)

(d)

(e)
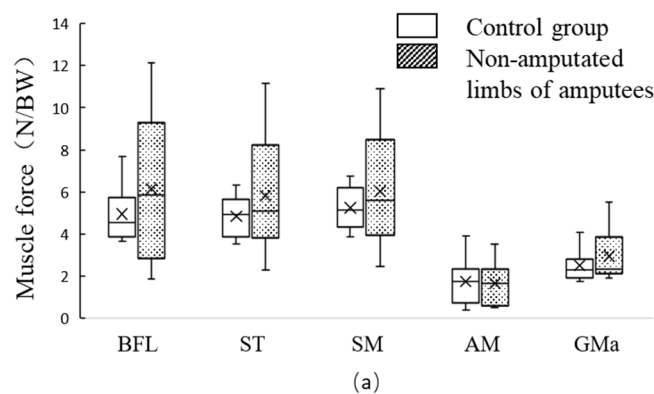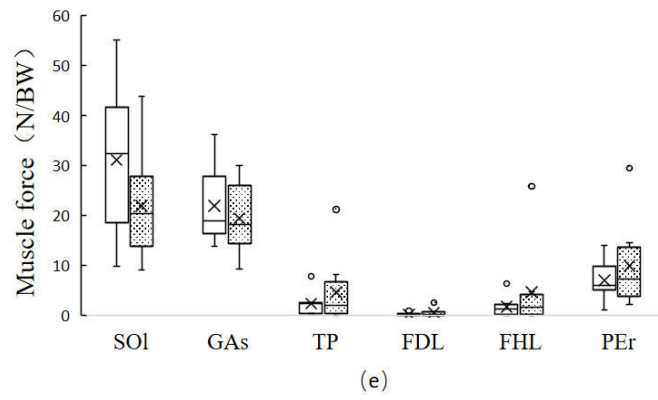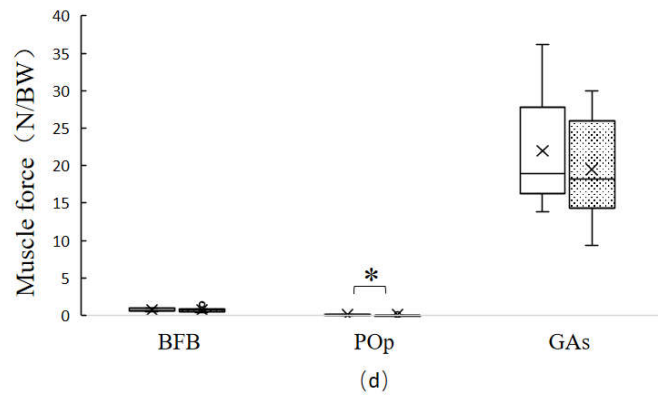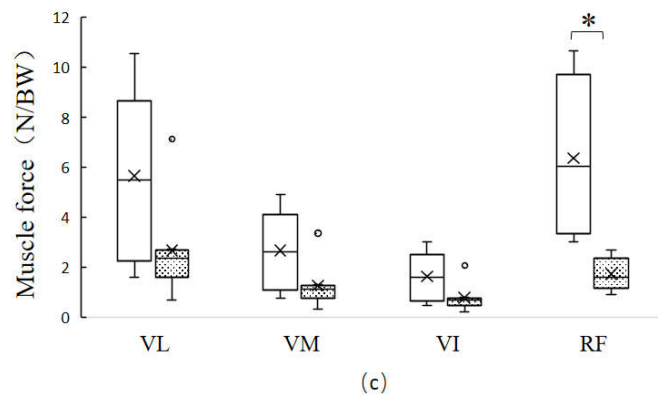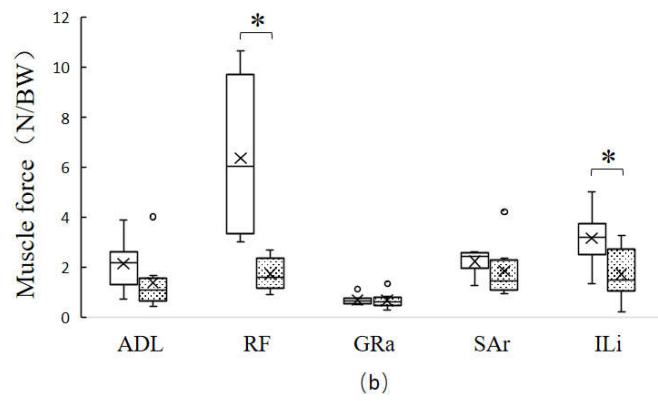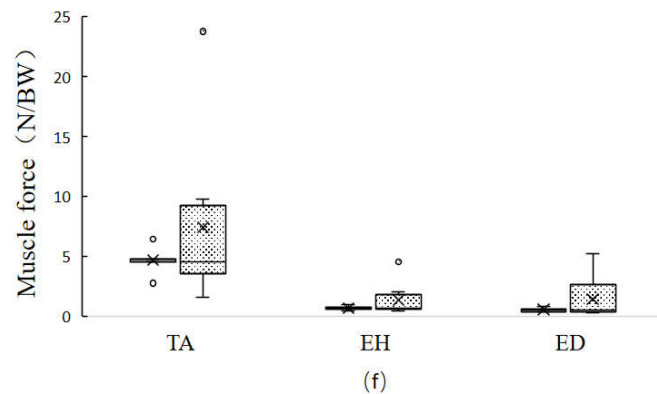
Fig. 7-13 Maximum force of each muscle: (a) hip extensors; (b) hip flexors; (c) knee extensors; (d) knee flexors; (e) ankle plantarflexors; (f) ankle dorsiflexors. ☆ indicates that there is a significant difference between the non-amputated limb of amputees and the control group

The POp in the knee flexors of non-amputated limbs of amputees was also reduced by 55.4% (p = 0.02), only amounting to 0.03 ± 0.03 N/BW, compared to control group. In addition, the statistical difference p values of the VL, VM, and VI of non-amputated limbs of amputees and the control group were of approximately 0.07 respectively, while the mean difference was approximately 52.6%.

The research presented in this chapter analyzed the muscle coordination of non-amputated limbs of unilateral transfemoral amputees and compared the results to abled persons. The muscle coordination was investigated in conjunction with joint moment. The results showed that the muscle coordination of non-amputated limbs was found to vary greatly from abled people. An investigation into muscle coordination of non-amputated limbs of transfemoral amputees may help to facilitate the improvement of rehabilitation and prosthesis design.

7.2 Joint moment and muscle coordination

This research investigated the characteristic instants of hip, knee, and ankle, i.e., H1 and H2 instant for hip joint, K1 and K2 for knee joint, and A1 and A2 for ankle joint, as shown in Fig. 6-1, Fig. 7-2, and Fig. 7-3.

While the joint moment of amputees presents no significant difference from that of the control group at H1, A1, and A2 instants, there is no difference in muscle force for the related joint muscles, as shown in Fig. 7-10 a, Fig. 7-12 a, and Fig. 7-12 b. Thus, when the joint moments are similar, the amputees' muscle coordination is similar to that of abled persons. However, the change in the required joint moment in gait is the motivation of the muscle coordination change.

At instant H2, the moment of the hip joint of amputees was approximately 58% smaller than that of the control group, while there was no significant difference in the joint angle. The difference in hip moment was attributed to the requirements of hip angular velocity. Although it can be seen from Fig. 7-1 a that the trajectory of non-amputated limbs joint angle curve of amputees and that of control group is relatively similar in the whole gait cycle. However, it is worth noting that the joint angle was usually a comparison scaled to 100% of the gait cycle. The duration of the gait cycle of amputees lasted longer, and the proportion of the stance phase to the whole gait cycle was also longer. Thus, the duration of the stance phase of non-amputated limbs is longer than that of the control group. H2 occurred in the pre-swing phase, which started with hip flex after it reached the maximum extension angle, and was ready to enter the swing phase. The moment of hip joint was affected by angular acceleration. The motion of hip joint changed from joint extension (the

angular velocity is negative) to joint flexion (the angular velocity is positive) at H2. The hip joint angular velocity of the control group showed a faster flexion speed, and the acceleration of angular velocity was also faster than that of non-amputated limbs because of the moment of inertia, and thus, the joint moment of the control group was greater than that of non-amputated limbs.

At H2 instant, the joint moment is different between the two groups, and the muscle forces of the hip joint were also significantly different. Except for the GRa, the muscle force of the other four hip flexors showed a decrease of 49%–81% compared with the control group, as shown in Fig. 7-10 b. The reduction in muscle force of the hip flexors directly caused the reduction in hip flexion moment. Therefore, the change in muscle activity of hip flexors directly contributed to the different hip moment of amputees. However, in order to achieve the specific hip moment required for the prosthetic gait, not all the muscle forces of the hip flexors of non-amputated limbs were reduced. The function of the GRa is not only to flex the hip joint, but also to adduct and medially rotate the hip joint. At H2, the forces of the GRa of non-amputated limbs of amputees were kept at a similar level to that of abled persons, while contributed little to the adjustment of hip moment. Although the muscle forces of the other four hip flexors were reduced to the lower hip moment, the GRa of the non-amputated limbs kept an activity similar to that of abled persons to maintain the other function in this circumstance, namely, to adduct and medially rotate the hip joint. In addition, the changes in hip muscle coordination were also in accordance with what we mentioned in Chapter IV. In that chapter we found that the intact hip joint plays a vital role in gait stability adjustment.

The knee joint of the control group showed a higher extension moment at K1, while

the knee moment of amputees was very small, with an average of only 0.04 ± 0.32 Nm/BW. At this instant, the muscle force of the quadriceps, except the RF, in the non-amputated limbs of the amputees showed a 64% decrease compared with that of the control group, as shown in Fig. 7-11 a. The decrease in their muscle force directly contributes to the decrease in knee moment of non-amputated limbs. At the same time, the muscle force of the BFB of non-amputated limbs increased, which contributed to the reduction of the knee extension moment as an antagonist muscle. However, the muscle force of the other knee flexors of non-amputated limbs, the POp, decreased. The POp was characterized by the function of flexing the knee joint and rotating the tibia inward. Besides, the reduction of the POp enabled the stiffness of the knee ankle to be reduced for its function as an antagonist muscle(Inoue et al., 2011). At K1, the muscle forces of the RF and GAs of non-amputated limbs were similar to those of the control group, and these two muscles did not contribute to the reduced requirements of joint moment. Both the RF and GAs belong to bi-articular muscles(Gj et al., 1987; Jacobs et al., 1996), in which the RF is defined as both a hip flexor and knee extensor, and the GAs is defined as both a knee flexor and ankle plantarflexor. Changes in their muscle forces affect multiple joints simultaneously. For this reason, RF and GAs of non-amputated limbs maintain similar muscle forces in abled people and contribute little to the specific knee moment requirement at the K1 instant.

The joint moment required for the prosthetic gait of amputees is different from that of abled persons, and the muscle force is essential for driving and controlling joints and producing the joint moment. From the muscle coordination of the hip joint at H2 and of the knee joint at K1, it can be observed that the muscle coordination of

amputees' non-amputated limbs has changed to meet the specific joint moment requirements. However, to achieve an appropriate joint moment, there are many possible combinations of muscle force. To achieve the different joint's moment requirements at the same time, it is necessary that all muscles coordinate with each other. Each muscle has its own unique function, and the muscle coordination adjustment is more complicated. For example, to achieve the purpose of reducing joint moment, not all joint flexors reduce their muscle force correspondingly. Certain muscles need to ensure a certain level of sitimulation to meet other specific functional requirements, such as the GRa of non-amputated limbs at H2, and the RF and GAs at K1. There may even be a combined action of both the active muscle and the antagonist muscle, such as in the knee joint muscles of non-amputated limbs at K1. As a result, when the requirement for joint moment changed, the adjustment of muscle coordination is one of the key factors to realize it. All the muscles will coordinate with each other while satisfying the basic requirements of gait to ensure gait stability, so as to adjust the joint moment and establish a new muscle coordination.

The different joint moment during gait is not always induced by muscle coordination alteration. The flexion moment of non-amputated limbs was much higher than that of the abled persons at K2, while the forces of knee flexors at this instant did not show any significant difference compared to abled persons. In fact, the muscle force of one knee extensor, RF, was significantly reduced. As discussed above, the RF is a bi-articular muscle whose activity affects both hip joint flexion and knee joint extension. The flexion moment of the hip joint of amputees was also significantly smaller than that of the abled persons at the K2 instant ($p<0.01$), as

shown in Fig. 7-1 b. Thus the difference in forces of RF between the two groups is attributed to the different hip moments.

The significantly increased knee flexion moment of non-amputated limbs has little relation to muscle coordination adjustment at the K2 instant; instead, the more anterior position of ground reaction force (GRF) induced the higher flexion moment. Further analysis shows that the vertical distance from the GRF vector to the centre of the knee joint rotation was 46.69±29.59 mm in non-amputated limbs at the K2 instant, much larger than 9.78±27.21 mm seen in abled persons (p=0.03). The greater the distance, the more anterior the GRF, producing higher external knee moment amplitude. K2 occurred in the terminal stance when the knee joint was at maximum extension. In order to balance the high external moment, the knee moment of non-amputated limbs was not realized by increasing the muscle force at the joint flexors but was passively balanced by knee joint fixation.

The hyperextension of the knee joint at K2 implicates secondary problems in non-amputated limbs. Several surveys have shown that non-amputated limbs of unilateral transfemoral amputees have a higher incidence of knee arthritis or knee joint pain than abled persons(Devan et al., 2012; Hagberg and Brånemark, 2001; Kulkarni et al., 1998; Norvell et al., 2005). E Hurwitz et al. believe that excessive force repeatedly acting on joints will cause pain and joint degeneration(E Hurwitz et al., 2001). Chang et al. suspected that joint pain in non-amputated limbs was related to long-term joint moment deviation(Chang et al., 2011), but their research only studied the joint adduction moment on the coronal plane. In fact, the motion range of knee joints in the sagittal plane is much larger than in the coronal or transverse planes, and it also produces a higher joint moment. As shown in Fig. 7-

2 b, non-amputated limbs of amputees demonstrated a moment of up to -0.44+-0.32 Nm/BW at K2, dramatically higher than seen in abled people. We believe that the pain or arthritis of knee joints in non-amputated limbs is related to the high joint moment at K2 in long-term daily life. In future rehabilitation processes, efforts should be taken to reduce the knee moment of non-amputated limbs at the K2 instant to relieve joint pain and osteoarthritis.

7.3 Maximum muscle force of flexors and extensors

It is commonly suggested that the muscles of non-amputated limbs of unilateral transfemoral amputees are likely undergo hypertrophy, but our study presents another possibility.

According to the analysis of the maximum value of each muscle force in the gait cycle, it can be observed that the maximum muscle force of non-amputated limbs was different from that of the control group. Compared with the control group, the RF, ILi, and POp of non-amputated limbs all showed a decrease in maximum muscle force, which was reduced by more than 45% for each one, as shown in Fig. 7-13. In addition, although the statistical p values of VL, VM, and VI were all approximately 0.07, they were beyond the significant difference threshold of 0.05. Considering the previous discussions showed that all three muscles reduced the stimulation level required to achieve a lower knee moment at the K1 instant, we believe the differences should not be ignored. The maximum forces of the three muscles also reduced by more than 50% compared to control group.

The duration of muscular activity of three thigh muscles in non-amputated limbs were discussed in Wentink's study (Wentink et al., 2013), and the effects of the

muscle force of four intact tibia muscles on the GRF were also discussed by Cerqueira et al. (2013), but maximum muscle forces were not reported in their studies. Sherk et al. found that the muscles of non-amputated limbs would atrophy during on-bed rehabilitation, while hypertrophy would occur when learning to restore bipedal locomotion ability (Sherk et al., 2010); Xiaolong Li also found that when an amputee started to wear prosthesis for rehabilitation training, the muscular volume of the non-amputated limbs appeared to have hypertrophy, while the muscular volume decreased after a period of rehabilitation (Li, 2016). He attributed the muscular hypertrophy to the overuse of the non-amputated limb. When an amputee started to learn how to walk with a prosthesis, it was a must for non-amputated limbs to undertake more work to compensate for the lack of mobility on the prosthetic limb, thus stimulating the muscles of non-amputated limbs and causing hypertrophy; However, with the adaptation to the prosthesis, the gait was more skillful and stable, and the compensatory gait was reduced, resulting in the reduction of the stimulation to the muscle and the muscular volume of non-amputated limbs. In spite of this, Xiaolong Li did not start tracking the muscular volume until one year after the amputation of the amputee. Sherk et al. started their research amputees of 19.7±11.1 years amputation history, and the muscular volume was compared to abled persons only, not with non-amputated limbs themselves during the process of rehabilitation. In addition, the muscular stimulation duration of the non-amputated limbs of the patient was longer (Castro et al., 2014; Cerqueira et al., 2013; Wentink et al., 2013), which may also induce muscle hypertrophy in rehabilitation. From this point of view, the lower muscle stimulation level during gait in long-term life may weaken the muscle function. Therefore, amputees should carry out pertinency training for the muscles during rehabilitation.

7.4 Summary

In this chapter, we analyzed the muscle coordination of non-amputated limbs of unilateral transfemoral amputees and compared the results to abled persons. A musculoskeletal model and dynamic simulation were utilized for calculating the muscle force. The muscle coordination of non-amputated limbs was found to vary greatly from abled people.

While there is no difference between the amputees and abled persons in joint moment, the muscle coordination of amputees' non-amputated limb is similar to that of abled persons. The alteration of muscle coordination is a positive adjustment control strategy for restoring locomotion ability that also impacts non-amputated limbs. In this process, the realignment of muscle coordination is relatively complex. The muscles will coordinate with each other and a new muscle coordination will be established. Nevertheless, the new muscle coordination needs to satisfy the other requirements to ensure gait stability and achieve the special joint moment requirement of a prosthetic gait.

The hyperextension of the knee joint at K2 implicates secondary problems in non-amputated limbs. The increase in flexion moment at this moment is less related to the adjustment in muscle coordination, but it is fixed to balance the extension moment of the knee joint caused by the GRF. We believe that the pain or arthritis of knee joints in non-amputated limbs is related to the high joint moment at K2 in long-term daily life.

Muscle disuse was found in non-amputated limbs. In the long-term daily life, if the maximum muscle force of the amputees has been at a low level for a long time, it

may cause the weakening of muscle function, and then affect the ability of the amputees to control their motions.

In general, the muscle coordination of non-amputated limbs varies greatly from abled persons. The change in muscle coordination indicates a change in motion control. An investigating into muscle coordination of non-amputated limbs is helpful for a deeper known about the motion control mechanism of amputees. The improvement of gait in an amputated limb cannot come at the expense of the non-amputated limb. The effects on the non-amputated limbs are required to be taken into account while improving the prosthetic product. In addition, the validation of the amputee's gaits cannot only be analyzed from the gait of the prosthetic limb, but also from the muscle coordination of non-amputated limbs, which should also be taken as an indicator of amputees' gaits.

CHAPTER VIII CONCLUSIONS AND SUGGESTIONS FOR FUTURE RESEARCH

There are numerous differences between the gait of transfemoral amputees and that of abled persons. Gait deviation exerts a serious impact on the amputees' health and daily life. Previous research on the gait of transfemoral amputees have underestimated the study of gait coordination, while the gait coordination is bound up with the rationality and health of amputees' gait. The insufficiency in theoretical guidance restricts the improvement of prosthetic products and the rehabilitation of amputees. The study of the gait coordination of amputees will help to shift the analysis of amputees' gait from single freedom, namely, each joint or each muscle separately, to multi-level freedom, from joints to joints and muscles to muscles. Hence, it can further analyze the gait characteristics of amputees from a new view and understand the weaknesses of amputees' motion control. Therefore, the rehabilitation programs and prosthesis design could be improved.

8.1 Major research achievements

In this study, gait coordination of unilateral transfemoral amputee was analyzed from multiple perspectives through gait analysis and inverse dynamics simulation.

The results show that the gait coordination analysis is more sensitive in detecting differences which are difficult to be found in the previous conventional gait analysis. In addition to pointing out the changes caused by the motion differences of individual joints, the inter-joint coordination analysis, which is one of the indicators of gait coordination, can also distinguish the changes of the coordination between

joints. In addition, we found that the hip joints of non-amputated limbs played a vital role in maintaining gait stability by adjusting the joint angular velocity and duration of swing phase. The ankle joint of a non-amputated limb prolonged duration of plantarflexion moment to enhance joint stiffness. The more rigid motion system reduced degree of freedom for motion control, which would reduce the difficulty of motion control and enhance motion steadiness.

The gait coordination of amputees was different from that of abled persons, which indicates that the motion control strategy was changed as their musculoskeletal system is different from that of abled persons. Nevertheless, similar coordination modes were found among the amputees. In addition, passive transfemoral prostheses, including then knee−ankle coupling designed prosthesis, have exerted consistent impacts on the gait coordination. Even if different amputees wore different types of prosthesis, a reasonable gait coordination variation was found among them. The consistency of effects of the prostheses on coordination indicates similar reliability in satisfying the basic functions. However, all the passive transfemoral prostheses showed that the degree of freedom of the prosthetic knee decreased in the stance phase, which affects gait coordination.

The knee–ankle coupling designed prosthesis has certain advantages in improving the gait coordination. In addition to the advantages of the linkage mechanisms found by previous studies, this study confirmed that the mechanism can also improve gait compliance of the amputated limbs when the center of gravity shifts forward during the stance phase.

The gait coordination of amputees varied with walking speed. The hip joints of both

amputated and non-amputated limb played a vital role in the adjustment of walking speed. Besides, the adjustment of the non-amputated limb in response to different walking speeds mainly occurred in the stance phase, which is similar to that of abled persons. The knee–ankle coordination of prostheses was less adaptability to different speeds, which should be strengthened in future designs. The speed at which amputees feel comfortable is the optimal speed for gait coordination. It is more flexible for amputees to control their joints at the speed. When walking at a slow speed, the amputees will have greater difficulty in controlling their motions and more challenges in coordination regulation. Amputees prefer to walk at a slow speed is the result of control adjustment to ensure an accurate and stable gait movement. The musculoskeletal system of the amputee can adjust the control strategy by slowing down the speed of motion and increasing the decomposition movement. It is a half measure to ensure gait stability. The slowdown of speed is merely a solution, and the inadequacy of amputees' ability to cope with motion control still needs to be improved. It is difficult for non-amputated limb joints of amputees to rotate at high angular velocity in gait, which directly affects the walking speed. The athletic ability of amputees' intact joints, such as the hip joint of a non-amputated limb, needs to be enhanced for improving the motion control ability. It is supposed to be beneficial to gait coordination of amputees directly.

The muscle coordination of non-amputated limbs was found to vary greatly from abled people. While there was no difference between the non-amputated limbs and the limbs of abled persons in joint moment, the muscle coordination of the two groups was similar too. Realignment of the muscle coordination was the main means to satisfy the specific joint moment requirements of non-amputated limbs.

However, it involves complex processes for the realignment of muscle coordination, as the muscles need to coordinate with each other to meet the requirements of joint moments, while ensuring gait stability. The non-amputated limb knee joint of amputee may produce much higher flexion moment at the end of the stance phase than that of abled persons. The increase in knee flexion moment is less related to the adjustment of muscle coordination, but it is fixed to balance the extension moment, namely, hyperextend. The high incidence of knee joint pain or arthritis should be associated with long-term exposure to such high joint moments. What is more, muscle disuse was found in the non-amputated limbs. It may cause the weakening of muscle function and the ability of motion control.

In summary, in this work, it has been found that there are many differences between transfemoral amputees and abled persons in gait coordination. The differences in coordination were derived from the different control strategies of the human body, which is the countermeasure to cope with the insufficient motion ability. Meanwhile, the differences in gait coordination also indicated the functional defects of the prostheses. The mechanism of the knee–ankle coupling designed prosthesis is beneficial to gait coordination, but there is still room for further improvement. The muscle coordination and joint moments of amputees' non-amputated limbs vary significantly from those of abled persons in gait, which embody the motion control adjustment. The different muscle coordination patterns of amputees implicate secondary problems and defects of the prosthesis. The study of gait coordination of transfemoral amputees enables us to have a better understanding of motion control in amputees. The more intuitive understanding of the deficiencies is helpful in improving rehabilitation programs and prosthesis design.

8.2 Innovation and significance of this study

This work studies the gait coordination of transfemoral amputees by means of gait analysis and inverse dynamics simulation, the following innovative results were presented:

The gait coordination pattern of transfemoral amputees was obtained for the first time. In this study, a new approach was utilized to investigate the gait deviation of amputees, namely, gait coordination analysis. The new method is more sensitive in detecting gait deviation compared with the previous traditional gait analysis approaches. This new method could be beneficial in detecting amputees' gait deviation, improving rehabilitation programs, and prosthesis design.

This study also proved that the knee–ankle coupling designed mechanism of a transfemoral prosthesis has its unique advantage in improving the gait coordination. The function of the knee–ankle coupling designed mechanism is closer to that of the musculoskeletal system of human beings. The linkage between a prosthetic knee and ankle realized the prosthetic ankle dorsiflexion and plantarflexion "actively," which is beneficial for conducting a smoother gait. Nevertheless, it was also found that the prosthetic knee maintains similar function to those of other passive prosthesis. There is still a room for improvement, especially in the knee extension function, to induce better gait coordination.

The effect of walking speed on the amputees' gait coordination was clarified in this study. We found the different strategies of gait coordination when the amputees induced different walking speeds. The current transfemoral prostheses are adjustable in response to various walking speeds, and an investigation of the gait

coordination at different speeds helps us to know better the defects of the prostheses. In addition, we also found the essential role of the hip joint in coping with different walking speeds. Accordingly, the clinical rehabilitation could enhance the training of the hip joint.

Non-amputated limb muscle coordination of unilateral transfemoral amputees was presented. In this study, it was found that the realignment of muscle coordination pattern was the main means to satisfy the specific joint moment requirement of non-amputated limbs. Nevertheless, the different patterns of muscle coordination compared to abled persons also implicated secondary problems, such as knee osteoarthritis and muscle disuse. Understanding of the causes of secondary problems is beneficial for clinical diagnose and to distinguish functional defects.

This study investigated the gait coordination of transfemoral amputees. The gait coordination deviation of amputees embodies deficiencies in motion control ability and defects of prostheses. The gait coordination analysis demonstrated advantages in its multi-level research and it is more sensitive in detecting gait deviation. The outcomes will provide a new theoretical basis for prosthetic design and clinical rehabilitation in the future.

8.3 Limitations

Due to difficulties in recruiting amputees who meet the criteria, the small sample size in this study is a limitation. Hence, it restricted the study for further discovery. Moreover, only male participants were involved in this study.

Only passive prostheses were involved in this study. Nowadays, the active

prosthesis is indeed an interesting topic. Many researchers are working on it. An active prosthesis has certain advantages; however, we decided to investigated the passive prosthesis for reasons. First, the passive prosthesis is much more popular all over the world. It is cheap, the structure is simple, the weight is light, and the maintenance is easy. On the contrary, the active prosthesis is much more expensive that not every amputee is able to afford one. It is supposed that the passive prosthesis would still be the main tool for transfemoral amputees in the next decades, especially in the developing country or rural area. The second reason is that although we investigated the passive prosthesis, it does not mean the findings of this study are only generalized to passive prosthesis because the basic theory is the same substantially.

The traditional passive prostheses used in this study were not unified. The amputees were instructed to walk with their own daily used prostheses in the gait analysis experiment. Different brand of prosthesis has different effect on the prosthetic gait. The purpose of gait analysis A was to investigate whether this type of prosthesis, namely the traditional passive transfemoral prosthesis, had similarity in the effect on the gait coordination. For this reason, the differences between the prostheses were not discussed in this study. Their individual effects on gait coordination were expected to be examined in future study.

The model used in the inverse dynamics analysis was a general human musculoskeletal model, and the accuracy of the muscle forces estimated from the simulation might be challenged. Although the model used in this study was scaled according to the anthropometric measurements of each subject, individualized muscle morphology and properties were not considered and the non-amputated

limbs were regarded as having the same musculoskeletal system as an able-bodied person's. Work is required by employing other control solutions, such as an EMG-informed approach, with more precise subject-specific physiological model to gain more in-depth understanding of the non-amputated limb muscle coordination.

In this study, the discussion of gait analysis was only in the sagittal plane. The inter-joint coordination in the coronal and transverse planes is worth studying in the future.

8.4 Directions of future studies

Based on this study, the following approach can be implemented for further improvement:

1) Further study on the gait coordination of transfemoral amputees with a larger sample size to collect more statistical results. Besides that, female amputees should be involved.

2) Investigating gait coordination wearing powered active prostheses, which can be meaningful to evaluate the rationality of the prostheses.

3) The gait coordination features of the prosthetic gaits are expected to be determined for each brand of prosthesis.

4) Investigating gait coordination in the coronal and transverse planes to have a more comprehensive profile of the gait of amputees.

5) Improving the rehabilitation programs based on the performance of gait coordination for each amputee.

6) Measuring the metabolic cost rates of wearing the knee–ankle coupling designed prosthesis and compared to that of traditional passive prosthesis to

check whether knee–ankle coupling designed mechanism can reduce the users'

efforts while walking.

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