Robotica: (2019) volume 37, pp. 2035–2055. © Cambridge University Press 2019
doi:10.1017/S0263574719000663

Design and Experimental Evaluation of Wearable Lower Extremity Exoskeleton with Gait Self-adaptivity

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(Accepted 08 April 2019. First published online: May 21, 2019)

SUMMARY
In this paper, we present a passive lower extremity exoskeleton with a simple structure and a light weight. The exoskeleton does not require any external energy source and can achieve energy transfer only by human body's own gravity. The exoskeleton is self-adaptive to human gait to achieve basic matching therewith. During walking, pulling forces are generated through Bowden cables by pressing plantar power output devices by feet, and the forces are transmitted to the exoskeleton through a crank-slider mechanism to enable the exoskeleton to provide torques for the ankle and knee joints as required by the human body during the stance phase and the swing phase. Our self-developed gait detection system is used to perform experiments on kinematics, dynamics and metabolic cost during walking of the human body wearing the exoskeleton in different states. The experimental results show that the exoskeleton has the greatest influence on motion of the ankle joint and has the least influence on hip joint. With the increase in elastic coefficient of the spring, the torques generated at the joints by the exoskeleton increase. When walking with wearing k3EF exoskeleton at a speed of 0.5 m/s, it can save the most metabolic cost, reaching 13.63%.

KEYWORDS: Lower extremity exoskeleton; Gait self-adaptivity; Passive; Mechanical design; Experimental evaluation.

1. Introduction

In recent years, researchers have developed a variety of lower extremity exoskeletons, and the applications thereof are mainly divided into three categories: exercise enhancement, rehabilitation training and exercise assistance. Some researchers have developed exoskeletons that are used in the field of medical rehabilitation to enable patients to recover their walking ability, such as HAL,[1] Ekso[2] and ReWalk.[3] In order to achieve rehabilitation training for stroke patients, some researchers have applied functional electrical stimulation (FES) to lower extremity exoskeleton.[4] These

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exoskeletons adopt rigid structures and are relatively heavy and have large inertial forces. Many studies have shown that weight and inertia are important issues for lower extremity exoskeletons used for walking assistance and gait rehabilitation.[5,6] The development of flexible lower extremity exoskeletons overcomes some deficiencies of rigid exoskeletons. Harvard University developed two Exosuit exoskeletons[7,8] that are lightweight, soft and comfortable to wear. The exoskeletons can generate assistive torques at hip, knee and ankle joints so as to reduce metabolic cost of the wearers during walking. Kawamura et al.[9] proposed an orthosis designed for people with small gait disorders (such as the elderly) which uses straight fiber artificial muscles as actuators to assist the flexion of the hip joints during the swing phase. A recently developed kind of shorts[10] made of polyvinyl chloride (PVC) gel actuators achieves shortening and stretching by controlling power-on and discharging of the PVC gel actuators so as to assist the flexion of the hip joints during the swing phase. In order to achieve better trajectory tracking, some researchers have also studied the control methods of pneumatic artificial muscle actuators.[11,12]

The common feature of these exoskeletons is that they all need an external energy source to provide power, and thus endurance is a problem. Therefore, some researchers have conducted research on passive lower extremity exoskeletons. Based on motion mechanism of human running, researchers have designed a passive exoskeleton that makes running easier.[13,14] There are also some exoskeletons that can reduce the gravity withstood by various joints of lower extremity when walking and increase the body's ability to withstand loads.[15–17] Dijk et al.[18,19] proposed a passive exoskeleton for reducing biological moment provided by joints when walking. The exoskeleton uses artificial tendons to store and release energy. However, experiments have shown that the exoskeleton cannot reduce the metabolic cost of the human body when walking. Collins et al.[20] designed an unpowered ankle exoskeleton. The exoskeleton uses a specially designed passive clutch to control locking and releasing of the spring. The clutch is locked when the foot touches the ground and lays flat. The spring is gradually stretched with the walking of the human body and contracts from the beginning of the pre-swing phase to produce a very large plantar flexion torque at the ankle joint. By testing the exoskeletons installed with different springs, it was found that the exoskeleton can save 7.2% of metabolic cost at most for the human body when walking.

Studies have shown that when the human body is in states of sit-to-stand, climbing stairs and upslope[21], the knee joint provides a positive power. For the elderly, with the deterioration of muscles, the walking ability declines, thus the flexion angle of the knee joint and the toe clearance decrease. All of these increase the falling risks when toes touch the ground during the swing phase. During walking, in order to ensure that the toes lift off the ground during the swing phase, the flexion angle of the knee joint is at least 60°.[22] When walking at a slow speed, the flexion speed of the knee joint decreases at the moment when the toes lift off the ground, which will also cause the maximum flexion angle of the knee joint to become smaller.[23] A large number of studies have shown that the human body needs the exoskeleton to provide a flexion torque to assist the flexion of the knee joint so as to achieve a sufficiently large toe clearance. For patients with knee injuries (such as knee arthritis, flexor injuries), maintaining the extension of the knee joint during the stance phase will further damage the soft tissues and muscles and will also cause a risk of falling because of failing to maintain extension due to pain. Therefore, the human body also needs the exoskeleton to provide an extension toque to keep the body stable.

Since the exoskeleton adopting a passive design may minimize its self-weight and is more portable and continuously used for a long time, we intend to develop a passive exoskeleton that can provide assistance to multiple joints and significantly reduce the metabolic cost of the human body when walking. The exoskeleton can automatically adjust according to the gait law of the human body to provide assistive torques to the knee and the ankle joint at an appropriate time during the stance phase and the swing phase. The exoskeleton can not only compensate for the insufficient exercise capacity of the patients and the elderly to enable them to restore normal walking ability, but it also can provide assistance for the healthy people to reduce muscle activities and save metabolic cost when walking.

2. Mechanical Design of Exoskeleton

2.1. Design ideas

The study of the gait of human body shows that a gait cycle can be divided into eight phases: initial contact, the loading response phase, the midstance phase, the terminal stance phase, the pre-swing phase, the initial swing phase, the mid-swing phase and the late-swing phase.[24] In order to match

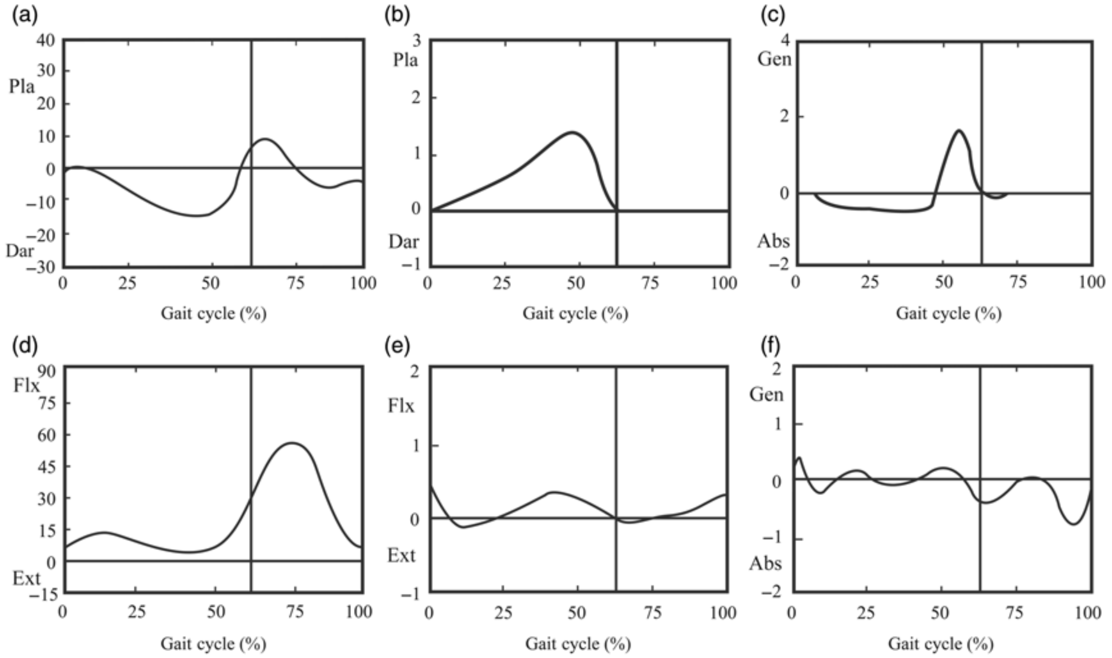

Fig. 1. Curves of ankle and knee joints: (a) ankle angle; (b) ankle moment; (c) ankle power; (d) knee angle; (e) knee moment; (f) knee power.

the motion law of the designed exoskeleton with the gait of the human body, it is necessary to firstly analyze the motion law of each joint during the walking of the human body. During the walking of human body, the main motion of the lower extremity joints is concentrated in a sagittal plane, so the research work on the joint motion is performed in the sagittal plane. Due to different test subjects and test methods chosen by different researchers, the resulting gait parameters are different, but the basic law and trend are identical. We use the joint angle, moment and power data provided by Kirtley in the CGA for analysis,[25] as shown in Fig. 1.

It can be seen from the curve of ankle joint in Fig. 1 that the ankle joint provides a plantar flexion moment during 0–60% of the gait cycle that reaches a maximum of 1.6 Nm/kg at about 50% of the gait cycle. From the initial contact through about 45% of the gait cycle, the shank muscles (gastrocnemius muscle, soleus muscles, etc.) contract eccentrically to absorb energy and do negative work. Starting from 50% of the gait cycle, the shank muscles contract concentrically to burst with a large energy to do positive work.[26,27] At 55% of the gait cycle, the power reaches a maximum of 1.7 W/kg. By comparing the power curves of the joints of the lower extremity, it can be found that the power of the ankle joint is the greatest and is several times that of other joints.

The curve of the knee joint in Fig. 1 shows that the motion angle of the knee joint during the entire gait cycle is 0°–60°. At about 20–60% of the gait cycle, the knee joint provides a flexion moment that peaks at 0.25 Nm/kg at 45% of the gait cycle. At 50% of the gait cycle, the positive work done by the knee joint reaches a maximum of 0.2 W/kg, which generates a positive power to assist the flexion of the knee joint. Starting from 50% of the gait cycle, hamstring muscles provide a flexion moment for flexing the knee joint so that the flexion angle of the knee joint finally reaches 60°.[28]

It can be seen that about 50% of the gait cycle is a critical time point through a comprehensive analysis of the curve of the ankle joint and knee joint. At this time, both the ankle joint plantar flexion moment and the knee joint flexion moment reach the maximum values. Therefore, when designing the exoskeleton, making the exoskeleton provide moments with the same trend to the ankle and knee joints at this time can maximize the assistance effect. Due to the gravity of the body, a great impact is generated when the foot initially touches the ground. The leg is subjected to inertial load brought by the body so as to decelerate the body and absorb the energy to keep the leg stable. This part of energy is lost in vain, and the state of the contralateral leg is 50% of the gait cycle. Therefore, a power output device installed on the sole of the foot is designed to transfer the energy generated when the foot falls down to touch the ground to the contralateral leg, so that the joints of the contralateral leg

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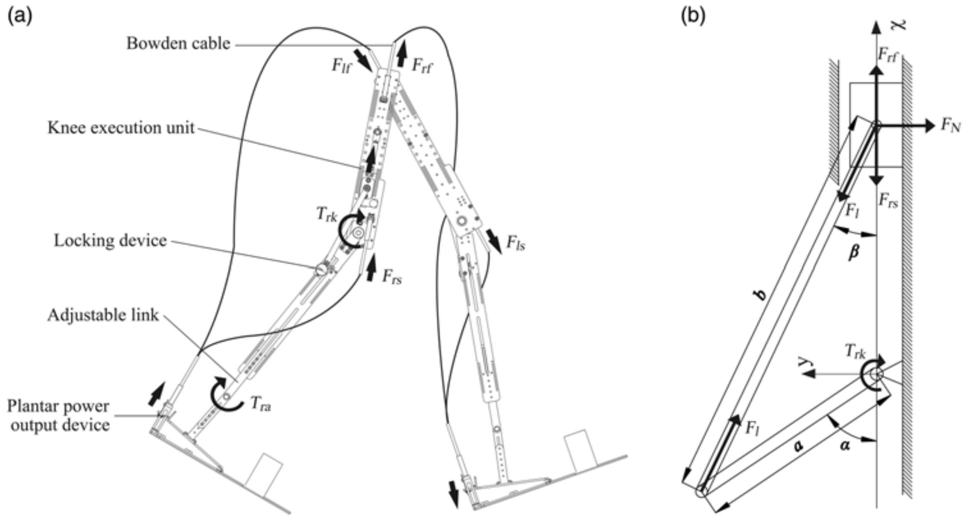

Fig. 2. (a) Overall force analysis of the exoskeleton. (b) Force analysis of crank-slider mechanism.

generate an assistive torque at 50% of the gait cycle. Finally the metabolic cost of the human body during walking can be reduced by sufficient use of the energy. Since the change in the joint biological moment is approximately linear, a spring can be used as an energy storage element.

2.2. Structure design

As shown in Fig. 2(a), the exoskeleton mainly consists of knee execution units, plantar power output devices, locking devices and Bowden cables. The knee execution units are fixed to the outside of two legs by straps. The plantar power output devices are small in size and installed on the heel, similar to the heels of a shoe.

When the locking device is locked, the slider transmits motion through the link to make the shank fixation plate rotate. When the left foot initially touches the ground, the upper plantar plate of the left foot is subjected to a force to move downwards so as to cause inner ropes of two Bowden cables to generate pulling forces. The force on the inner rope of the Bowden cable for flexion of the right leg is transmitted to the right shank fixation plate through the crank-slider mechanism so as to generate a flexion torque at the knee joint. Forces on the inner rope of the Bowden cable for extension of the right leg and the Bowden cable for flexion of the left leg form a plantar flexion torque at the right ankle, as shown in Fig. 2(a).

When the locking device is released, there is no correlation between the motion of the slider and shank fixation plate. During the walking of the human body, the exoskeleton does not exert any force on the knee joint, and forces are only transmitted between plantar power output devices on two sides through the Bowden cables. In this state, the plantar power output device on one side is pressed when the foot on this side falls down to the ground, which causes the upper plantar plate of the plantar power output device on the other side to move upwards by the pulling force of the Bowden cables and generate a plantar flexion torque at the ankle joint.

Due to the small size and light weight of the slider and the link, the force analysis of the crank-slider mechanism in the knee execution units of right leg is performed by ignoring the mass and friction of the slider and the link; see Fig. 2(b).

Assuming $F_{rf} > F_{rs}$, the following equations can be obtained according to force balance of the slider:

$$\begin{cases} F_l \sin\beta = F_N \\ F_l \cos\beta = F_{rf} - F_{rs} \end{cases} \tag{1}$$

where F_l represents the force on the link, and F_N represents the support force on slider. β is the angle between the link and the guide rail, F_{rf} represents the force on the inner rope of the Bowden cable for flexion of the right leg, F_{rs} represents the force on the inner rope of the Bowden cable for extension

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of the right leg and F_{ls} represents the force on the inner rope of the Bowden cable for extension of the left leg.

$$a\ \sin\alpha = b\ \sin\ \beta \tag{2}$$

$$b^2 - b^2\cos^2\beta = a^2\sin^2\alpha \tag{3}$$

where α is the angle between the crank and the guide rail, a represents the length of the crank and b represents the length of the link. The following equation can be obtained through solving Eqs. (1) and (3):

$$F_l = \frac{(F_{rf} - F_{rs})b}{\sqrt{b^2 - a^2\sin^2\alpha}} \tag{4}$$

It can be known by the force analysis for the link that

$$T_{rk} = F_l\ \sin\left(\alpha - \beta\right)\ a \tag{5}$$

The torque generated at the knee joint by the exoskeleton is obtained through solving Eqs. (4) and (5):

$$T_{rk} = \frac{(F_{rf} - F_{rs})ab\ \sin\left(\alpha - \beta\right)}{\sqrt{b^2 - a^2\sin^2\alpha}} \tag{6}$$

Eliminate β by replacing it with α and perform simplification to obtain the following equation:

$$T_{rk} = (F_{rf} - F_{rs})a\ \sin\alpha\ - \frac{(F_{rf} - F_{rs})a^2\sin\alpha\ \cos\alpha}{\sqrt{b^2 - a^2\sin^2\alpha}} \tag{7}$$

where T_{rk} represents knee torque of right leg. When $F_{rf} > F_{rs}$ and $T_{rk} > 0$, the exoskeleton provides a flexion torque to the knee joint; when $F_{rf} < F_{rs}$ and $T_{rk} < 0$, the exoskeleton provides an extension torque.

The following equation can be easily obtained according to the positional relation of the crank-slider mechanism:

$$x_s = \sqrt{b^2 - a^2\sin^2\alpha} - a\ \cos\alpha\ \ - x_i \tag{8}$$

where x_s represents the displacement of the slider and x_i represents the distance between the slider and the center of rotation when the thigh plate and the shank plate are in horizontal, that is, the initial position of the slider.

The torque at the ankle joint is generated by the pulling forces on the inner ropes of the Bowden cables for flexion and extension. The force arm at the ankle d is obtained by measuring the vertical distance from the center of the ankle joint to the pulling force on the inner rope. Ultimately, the torque generated at the ankle joint by the exoskeleton is as given below:

$$T_{ra} = (F_{rs} + F_{lf})d \tag{9}$$

where T_{ra} represents the torque at the ankle joint of the right leg and F_{lf} represents the force on the inner rope of the Bowden cable for flexion of the left leg.

a, b, x_i, α_i and d are all known parameters from mechanical structures, where α_i is the initial angle between the crank and the guide rail when the thigh plate and the shank plate are in horizontal. By substituting F_{rf}, F_{rs}, F_{lf} and α measured by the sensor and known parameters into Eqs. (7) and (9), the torques generated at the knee and ankle joints by the exoskeleton can be obtained.

When the locking device is locked, the execution process of the exoskeleton during the entire gait cycle[29] is shown in Fig. 3. Based on principles of lever and crank-slider mechanism, the exoskeleton utilizes the human body's own weight as power during walking. The plantar power output devices are alternatively pressed by both feet to cause the Bowden cables to generate pulling forces and finally drive the crank-slider mechanism to perform a cyclical reciprocating motion to generate torques at the ankle and knee joints. The exoskeleton can automatically adjust according to the gait law of the human body to assist flexion of the knee joint during the terminal stance phase, the pre-swing phase,

Table I. Mass of respective part of the exoskeleton.

Name of the part	**Mass (kg)**
Single knee execution unit	0.654
Single plantar power output device	0.494
Total mass	2.296

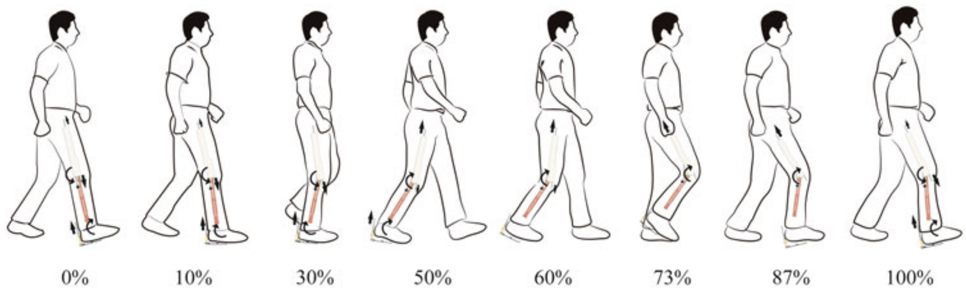

Fig. 3. Execution process of the exoskeleton during a gait cycle.

the mid-swing phase and the late-swing phase and assist plantar flexion of the ankle joint during the stance phase. In addition, it is possible to provide torques for assisting extension of the knee joints during the midstance phase. Through the transmission of the link mechanism formed by the lower extremities of the human body, a small torque is also generated at the hip joints. This paper mainly considers the moments formed at the knee and ankle joints of the human body by the exoskeleton.

In order to reduce the mass of the exoskeleton while ensuring its rigidity, the main material of the exoskeleton is aluminum alloy. The plantar plate is made of stainless steel to ensure that it will not deform during a long time walking of the human body. The mass of respective part of the exoskeleton is shown in Table I. The knee execution units and plantar power output devices of two legs are staggered and connected together by Bowden cables. The specific connection relation is shown in Fig. 4(b).

Figure 5 is a structure diagram of the plantar power output device. The plantar power output device mainly consists of a lower plantar plate, an upper plantar plate, a bracket and Bowden cables. The lower plantar plate and the upper plantar plate are connected by a low friction hinge and can achieve relative rotation with each other. The inner ropes of the Bowden cables are fixed with the upper plantar plate, and the outer covers thereof are against the bracket. Since the plantar power output device is installed at the heel, it must be ensured that the device will not come into contact with the ground during the swing phase. By studying the minimum clearance between the foot and the ground during the swing phase, the maximum displacement of the upper plantar plate is finally determined to be 0.04 m. In order to improve comfort so that people do not feel uneven under the foot, a rubber pad can be installed on the position of the lower plantar plate corresponding to the fore sole of the foot.

As shown in Fig. 6, the knee execution unit is mainly composed of a thigh fixation plate, a shank fixation plate, a crank-slider mechanism and Bowden cables. The left end of the slider is connected with the inner rope of the Bowden cable for flexion by a linear spring, and the other end of the Bowden cable is connected with the plantar power output device of the contralateral leg. The right end of the slider is directly connected with the Bowden cable for extension, and the other end of the Bowden cable is connected with the plantar power output device of the ipsilateral leg. By pressing the left and right plantar power output devices, the slider can be moved back and forth to achieve flexion motion and extension motion of the knee joint. When the displacement of the slider is 0.03 m and the spring is not elongated, the corresponding rotation angle of the shank fixation plate is 60°, and the exoskeleton provides assistive torque for the motion in this range. Since different people have different maximum knee flexion angles, in order to prevent the exoskeleton from hindering normal motion of the knee joint of the human body during walking, the maximum flexion angle that the exoskeleton can reach is designed as 70° and the exoskeleton does not provide assistive torque during the motion between 60° and 70°.

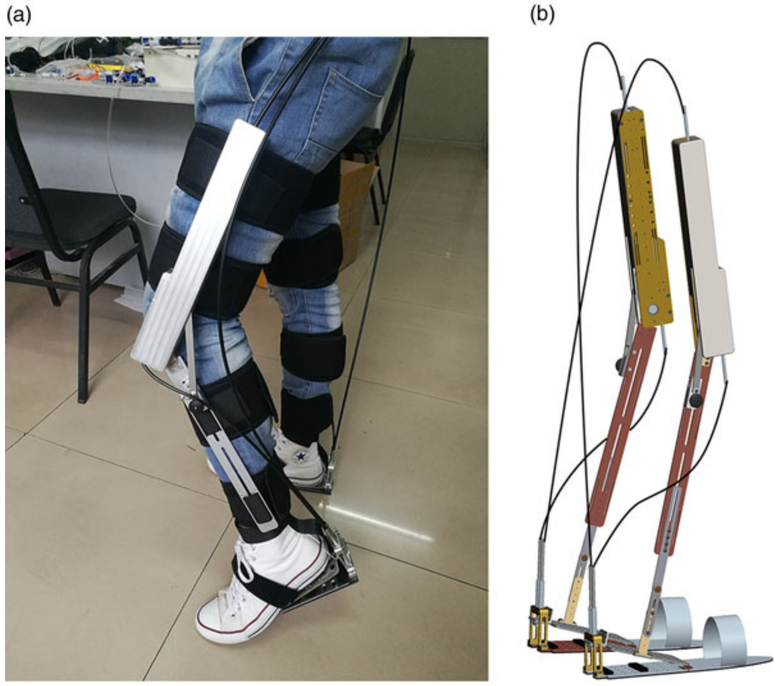

Fig. 4. (a) Photograph of human body wearing the exoskeleton. (b) Connection relation of Bowden cables between the exoskeletons.

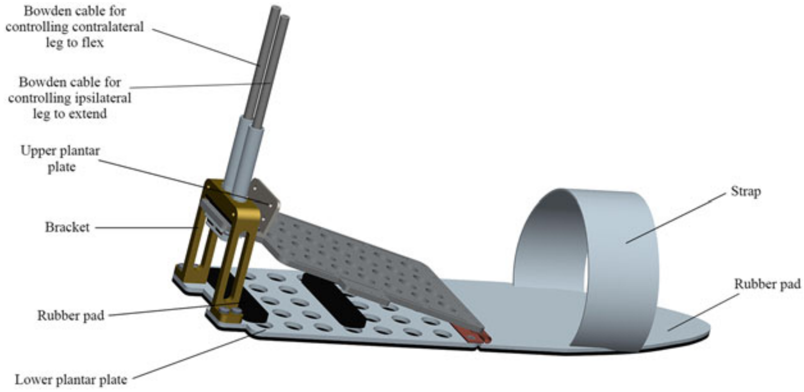

Fig. 5. Structure diagram of plantar power output device.

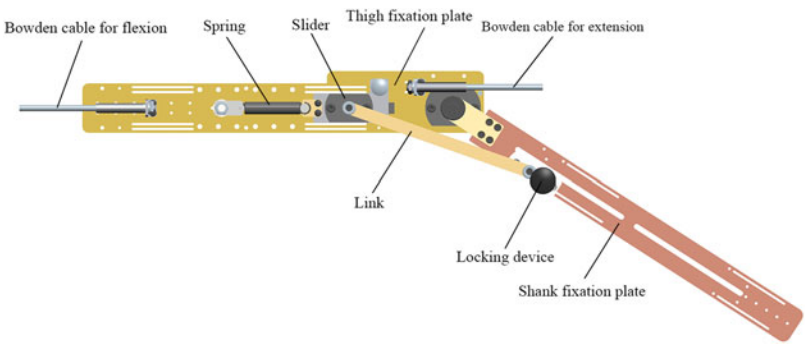

Fig. 6. Structure diagram of the knee execution unit.

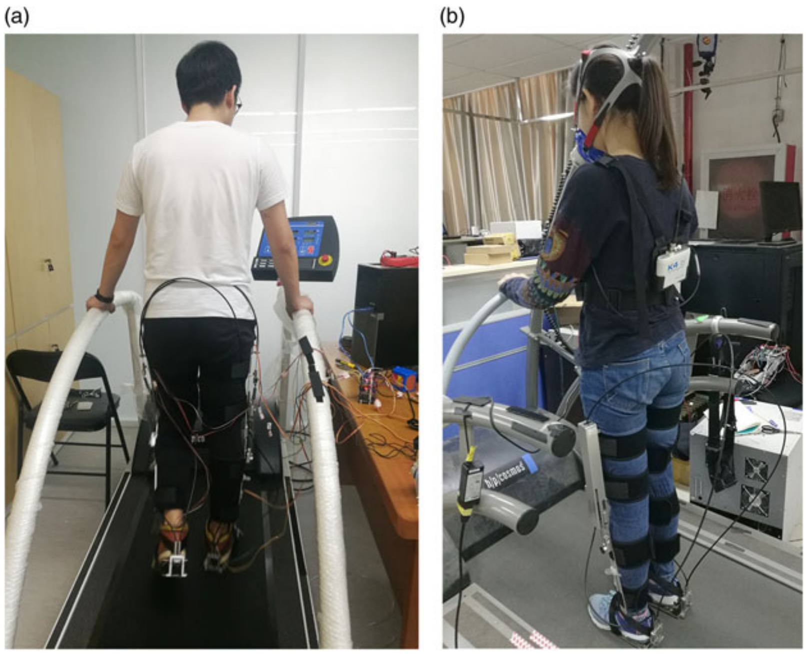

Fig. 7. (a) Photograph of kinematics and dynamics experiment. (b) Photograph of metabolic cost experiment.

The adjustable link connects the shank fixation plate with the plantar power output device through a spherical pair and makes the position of the spherical pair coincided with the ankle joint of the human body by adjusting the length thereof. In this way, it is ensured that the weight of the exoskeleton is transmitted to the ground through the plantar power output devices without influencing normal motion of the ankle joint, thereby reducing the burden on the leg of the human body caused by the self-weight of the exoskeleton.

The motion law of the exoskeleton in execution process is consistent with the gait law of the human body. Thus it will inevitably influence other motions of the lower extremities of the human body. In order to connect and disconnect the connection between the link and the shank fixation plate, a locking device is designed. After the locking device is released, the link and the shank fixation plate are no longer connected, and the knee joint can move freely without any restriction, and the human body can stand, sit-to-stand and squat at this time freely.

3. Experimental Methods

The contents for experiments mainly comprise experiments for kinematics, dynamics and metabolic cost, as shown in Fig. 7. The purpose of kinematic experiment is to analyze the influence of the exoskeleton on joints of lower extremities of the human body during walking. The dynamic experiment is used to evaluate the torque trends generated at various joints of the human body by different exoskeletons. The purpose of design for the exoskeleton is to save energy consumption of the human body during walking. Therefore, the metabolic cost experiment on the human body can verify the overall influence of the exoskeleton during walking. To facilitate experiment, we have developed a gait detection system that can measure angles of the hip, knee and ankle joints of the human body.

3.1. Experiment conditions

The purpose of design of the exoskeleton is to provide assistance for the walking of the elderly, the knee injuries and the healthy people and reduce energy consumption. To be on the safer side, we first choose to perform experiment on healthy people wearing the exoskeleton. Six subjects (three males, three females, average age was 25 (SD = 4) years, average height was 1.71 (SD = 0.1) m, average body mass was 60 (SD = 10) kg) participated in the metabolic cost experiment. The experiment contents include three ways: the human body does not wear the exoskeleton, the human body wears the exoskeleton without knee joint assistance and the human body wears the exoskeleton with knee joint assistance. The state of the exoskeleton without knee joint assistance is to release the locking

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Table II. Spring parameters used in the exoskeleton.

Code	**Elastic coefficient (N/m)**
k1	94
k2	904
k3	2041
k4	4224
k5	6022

device, the knee joint is no longer subjected to the assistive torque at this time and the ankle joint is subjected to the plantar flexion torque. The state of the exoskeleton with knee joint assistance is to fix the locking device, and the exoskeleton provides torques to the knee joints and provides plantar flexion torques to the ankle joints at this time. We use the abbreviation NW to represent not wearing the exoskeleton, NF to represent the exoskeleton without knee joint assistance and EF to present the exoskeleton with knee joint assistance.

When the same person walks at the same speed, the spring is the only factor for regulating the size of the assistive torque. Therefore, in order to understand the influence of the exoskeleton installed with different springs on the human body, we performed multiple sets of experiments on the exoskeleton. Details of the spring installed in the exoskeleton are shown in Table II.

3.2. Design of gait detection system

The commonly used human gait detection system is 3D motion analysis system which can detect motion parameters of the lower extremity joints in various directions during walking of the human body. However, the device uses many cameras to capture image information, needs to install many marker points on the surface of the human body and can be performed only indoors. In order to facilitate the study of the human gait in various complicated outdoor environments and evaluate the performance of the exoskeletons during long-term walking, we have developed a portable gait detection system.

The gait detection system mainly consists of five 9-degree-of-freedom inertial measurement units (IMUs), four force sensing resistors (FSRs) and a data acquisition board. The gait detection system can detect the angles of the joints of the human body in sagittal plane. We installed three IMUs on the outside of the thigh, the shank and the foot to detect the angles of the joints of the lower extremity of the human body. The other two IMUs are installed on the outside of thigh fixation plate and the shank fixation plate of the exoskeleton and are used to measure the angles of knee joint of the exoskeleton. This is because a relative displacement will occur between the exoskeleton and the human legs during the walking of the human body wearing the exoskeleton, so it is necessary to perform detection for the exoskeleton and the human body, respectively. The angles of the hip, knee and the ankle joints can be calculated by subtracting the direction vectors of two adjacent IMUs.

FSR exhibits a decrease in resistance with an increase in the force applied to the active surface. We fix the FSRs on the insoles to make plantar pressure insoles for detecting changes in the plantar pressure during walking of the human body. The positions of the FSRs on the insoles are shown in Fig. 8(b). Each phase of the gait cycle of the human body can be determined according to the feature of each gait phase in the gait cycle combined with the data collected by the IMUs and FSRs. For example, when the resistance of the FSR at the heel begins to decrease from infinity, it means that the heel begins to touch the ground.

In order to understand more clearly the size of the force to which the FSR is subjected during walking, we calibrated the FSR as shown in Fig. 9. The FSR is connected in series with a resistor R of 10.3 kΩ. The pressure is continuously applied to the FSR by moving the pressure gauge, and the voltage across R is collected at the same time. The resistance of the FSR under different pressures can be obtained by using Eq. (10). The change in the value of the voltage across the resistor R can also reflect the change in the force to which the FSR is subjected. When the voltage across R increases, it means that the pressure to which the FSR is subjected increases.

$$\mathrm{R}_{\mathrm{FSR}} = 5R/\mathrm{V}_R - R \tag{10}$$

where R_{FSR} represents the resistance of the FSR and V_R represents the voltage across the resistor R.

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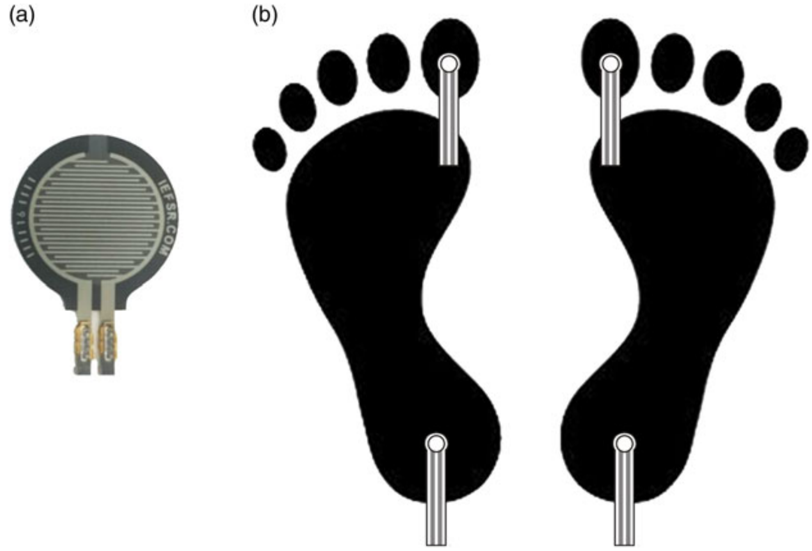

Fig. 8. (a) Photograph of force sensing resistors. (b) Positions of FSRs on insoles.

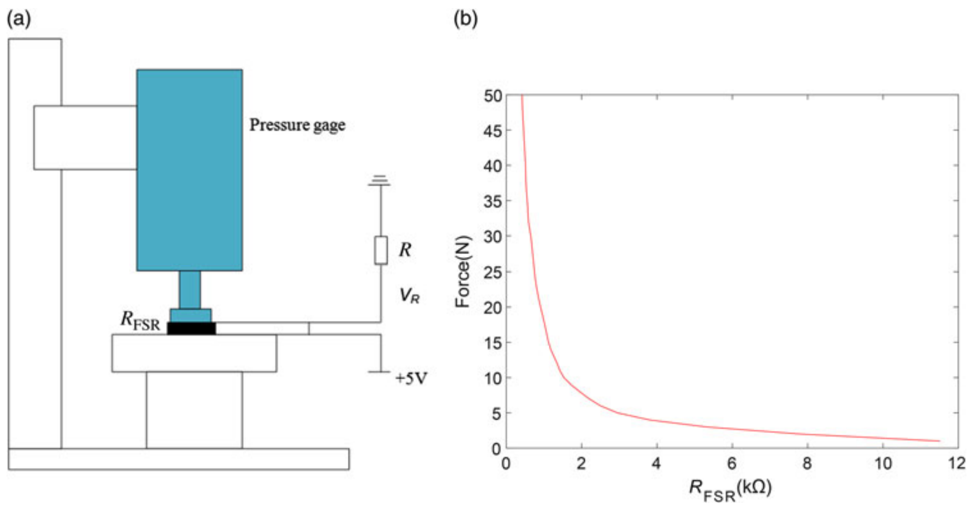

Fig. 9. (a) Schematic diagram of FSR pressure calibration. (b) Diagram of FSR resistance versus pressure to which it is subjected.

3.3. Kinematic experiment

Walking speed is one of the important factors influencing the gait parameters of joints of the lower extremity of the human body. The peak flexion moment of knee joint is also closely related to walking speed.[30] The elderly and the people with knee problems have slower walking speeds. Therefore, the angles of the hip, knee and ankle joints during walking of the human body at three speeds of 0.6, 0.8 and 1.0 m/s, respectively, wearing the exoskeleton are tested. The angles of each joint when standing still are used as zero, and the entire experiment process was performed on the H/P/COMSOS professional sport test treadmill.

3.4. Dynamic experiment

In order to calculate the torques generated at the knee and ankle joints by the exoskeleton, four miniature tension sensors are installed on the inner ropes of the Bowden cables. The torques can be calculated by bringing the detected pulling forces and angles into Eqs. (7) and (9). The torques generated at the joints by the exoskeleton are different when different springs are installed. In order to know the torque trends generated at each joint by the exoskeleton, the exoskeletons installed with five different springs were tested under three speeds of 0.6, 0.8 and 1.0 m/s.

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Because biological moment at the knee joint and gait phase of different people are different from each other. In order to make the exoskeleton adaptable to each individual, we hope to be able to adjust the period which the torque is provided by adjusting the angle range of the knee execution unit providing assistive torque. By adjusting the length of the inner ropes of the Bowden cables for flexion of the exoskeleton, the angle range of the knee execution unit providing assistive torque can be adjusted when the plantar power output devices are pressed. The torque trends generated at the joints by the exoskeleton are studied when the angle range of the knee execution unit providing assistive torque is 0°–40°.

3.5. Metabolic cost

In order to verify whether the exoskeleton can reduce the metabolic cost of the human body when walking, we used K4b^2 (COSMED, Rome, Italy) to test the human body. Because the exoskeleton is a passive exoskeleton, the size of the torque provided by the exoskeleton depends on elastic coefficient of the spring. Collins et al.[20] pointed out that the larger the elastic coefficient of the spring, the greater the torque generated. However, it is not true that the larger the elastic coefficient is selected, the more energy is saved for the human body, but the spring with a medium elastic coefficient is optimal. People tend to choose the pattern coordinated with net biological moment of the ankle joint.[31,32] So we think that the magnitude of the moment the exoskeleton produces at the joint does not indicate which exoskeleton is better. Therefore, the optimum parameters of the spring installed on the exoskeleton cannot be determined by kinematic and dynamic experiments. We consider that the kind of spring installed on exoskeleton which can save the most energy for human body in metabolic cost experiment is optimum. So the metabolic cost of the exoskeletons installed with five different springs was tested. The walking speed of the human body influences not only the gait match between the exoskeleton and the human body but also the metabolic cost of the human body walking wearing the exoskeleton. Therefore, the metabolic cost of the human body walking at six speeds of 0.5, 0.6, 0.7, 0.8, 0.9 and 1.0 m/s wearing the exoskeleton are tested.

During the experiment, each subject first stood still for 2 min and then walked for 5 min at each speed. The data taken from the stable walking during the middle 3 min of the walking were selected as valid data and the oxygen consumption VO_2 (ml/min) and carbon dioxide emission VCO_2 (ml/min) detected by K4b^2 during the walking of the human body were recorded. Metabolic cost of the human body was calculated using the Weir[33] formula. The formula is as follows:

$$EE = 3.9\mathrm{VO}_2 + 1.1\mathrm{VCO}_2 \tag{11}$$

where VO_2 (ml/min) represents O_2 consumption and VCO_2 (ml/min) represents CO_2 production. The resulting *EE* (cal/min) was converted in units and normalized by dividing by body weight (kg) to finally obtain Ekg (W/kg).

4. Experimental Results and Evaluation

The data in the kinematics and dynamics results are the average of multiple gait cycles. What needs to be specially explained is that since the plantar power output device has a certain size and thickness, the device has already touched the ground in advance before the human's heel initially touches the ground and a contact force is formed between the human body and the ground. By analyzing the joint angles and the plantar pressure data obtained from the gait detection, it can be found that the time span from the plantar power output device touching the ground to the heel touching the ground lasts about 10% length of the whole gait cycle. In terms of normal gait cycle 352 of the human body, the plantar power device of the exoskeleton comes into contact with the ground 353 at about remaining 10% of the late-swing phase.

4.1. Kinematics

By comparing the curves of NW and NF in Fig. 10, it can be seen that the angle of the hip joint changes little when walking at different speeds with wearing the NF exoskeleton, which indicates that the hip joint is less influenced by the exoskeleton. The maximum flexion angle of the knee joint during the swing phase is slightly reduced. The flexion angle of the knee joint increases when the foot initially touches the ground and the flexion angle during the stance phase also increases. With the increase in walking speed, the knee joint is affected more. It can be known by analysis that because the exoskeleton itself has a certain weight, this part of extra load influences the swing of

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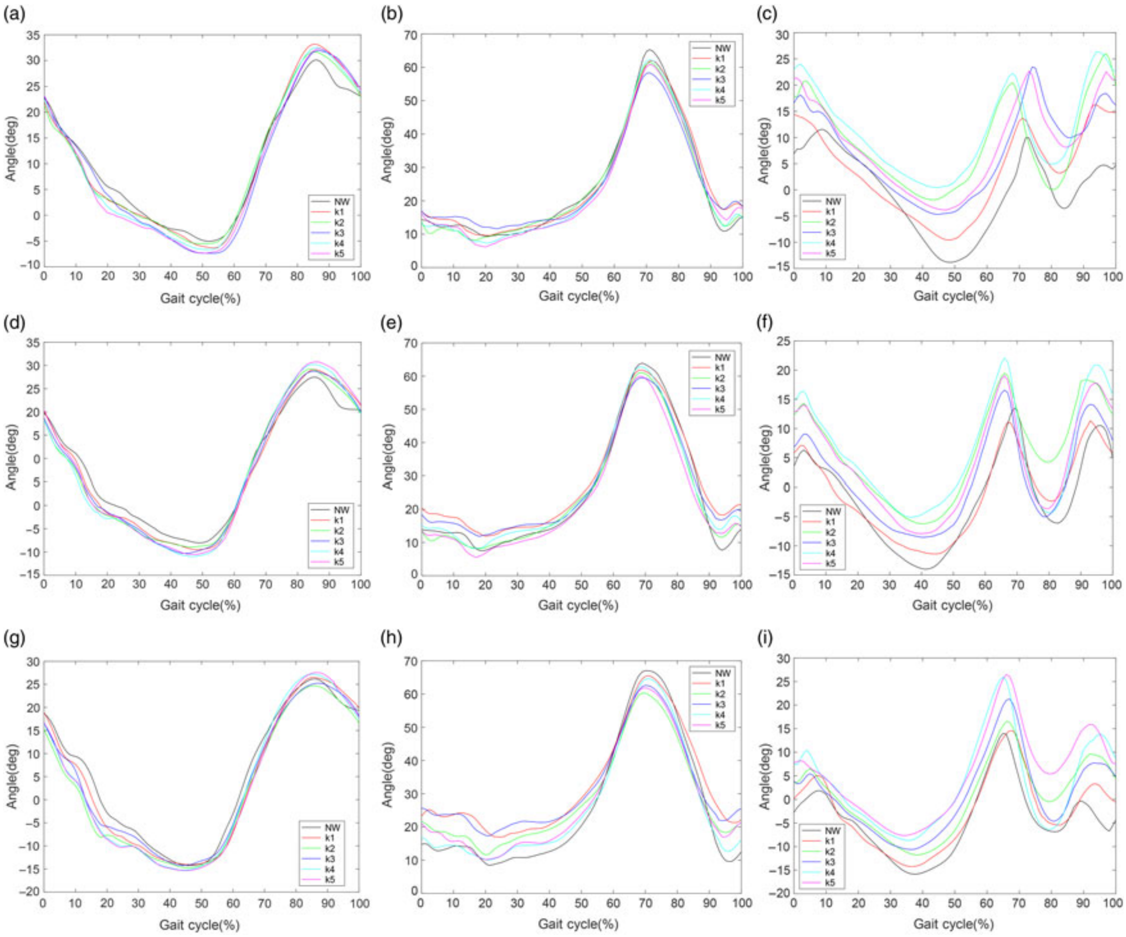

Fig. 10. Angles of lower extremity joints when walking horizontally, hip joint (left column), knee joint (middle column), ankle joint (right column); joint flexion and plantar flexion (>0), joint extension and dorsiflexion (<0): (a)–(c) show the occasion of walking at a speed of 0.6 m/s with wearing NF exoskeleton; (d)–(f) show the occasion of human walking at a speed of 0.8 m/s with wearing NF exoskeleton; (g)–(i) show the occasion of human walking at a speed of 1.0 m/s with wearing NF exoskeleton.

the lower extremity of the human body during walking, which causes the maximum flexion angle of knee joint in swing phase to be reduced. With the increase in the walking speed, the influence of the inertial force generated by the self-weight of the exoskeleton on the human body increases so that the maximum flexion angle of the knee joint during the swing phase is reduced. Starting from about 90% of the gait cycle, the flexion angle of the knee joint no longer decreases. This is because the plantar power output device comes into contact with the ground at this time which makes the leg fail to continue to complete the extension motion. The ankle joint is subjected to a greatest influence than other joints. Because the plantar power output device is installed at the heel of the foot, its own structure has a certain size and weight and the plantar flexion torque is generated during walking, which causes that the plantar flexion angles of the ankle joint increase and the dorsiflexion angles reduce during the entire gait cycle.

By comparing the curves of NW and EF in Fig. 11, it can be seen that the angle of the hip joint still changes little and is less influenced when walking at different speeds with wearing the EF exoskeleton. The flexion angle of the knee joint increases when the foot initially touches the ground, and the flexion angle during the stance phase also increases. With the increase in the walking speed, the maximum flexion angle of the knee joint during the swing phase decreases and the knee is affected more when walking with wearing the exoskeleton. However, the greater the elastic coefficient of the spring, the greater the maximum flexion angle of the knee joint during the swing phase. It can be known by analysis that the knee joint flexes during the swing phase. Meanwhile the knee joint is subjected to the flexion torque provided by the exoskeleton and also bears the load force brought

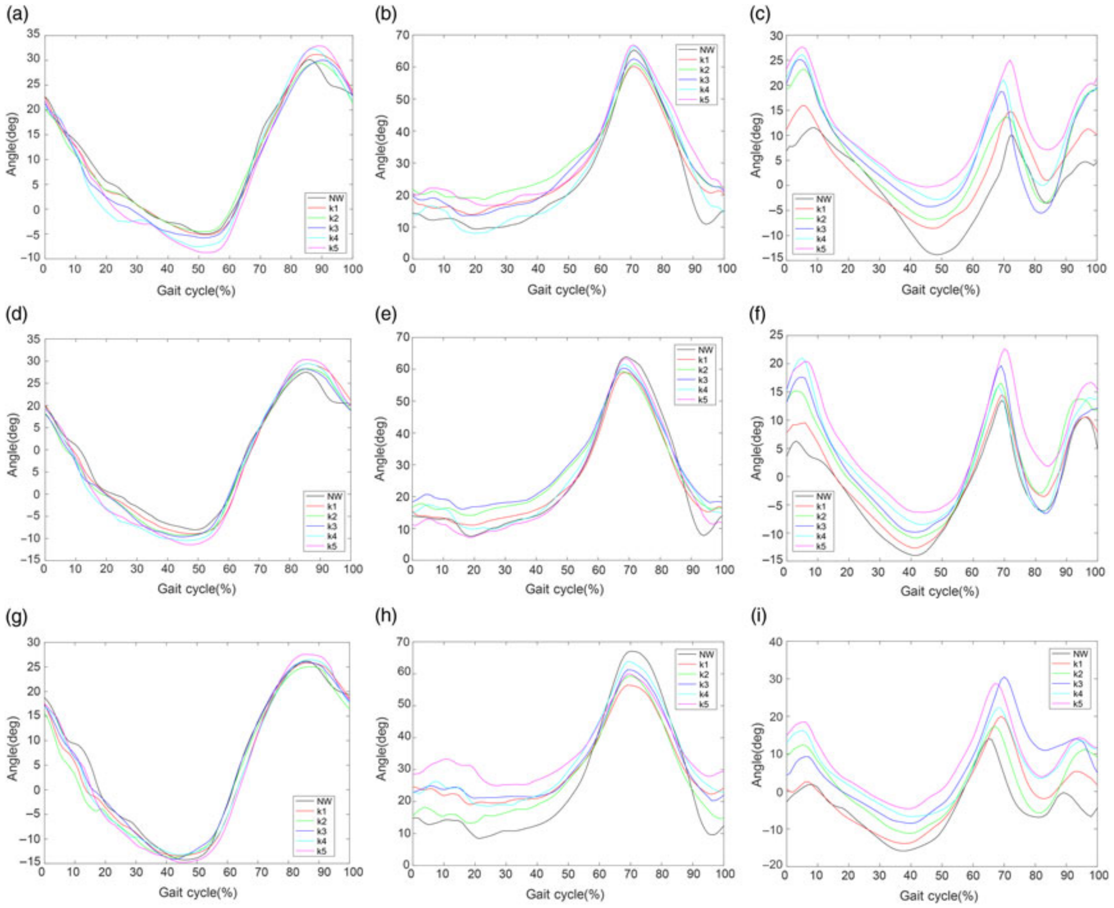

Fig. 11. Angles of lower extremity joints when walking horizontally, hip joint (left column), knee joint (middle column), ankle joint (right column); joint flexion and plantar flexion (>0), joint extension and dorsiflexion (<0): (a)–(c) show the occasion of walking at a speed of 0.6 m/s with wearing EF exoskeletons; (d)–(f) show the occasion of walking at a speed of 0.8 m/s with wearing EF exoskeletons; (g)–(i) show the occasion of walking at a speed of 1.0 m/s with wearing EF exoskeleton.

by the self-weight of the exoskeleton. When the elastic coefficient of the spring is large enough to eliminate the influence of the load force, the flexion angle of the knee joint increases. Otherwise, the flexion angle of the knee joint decreases. However, with the increase in walking speed, the influence of the inertial force generated by the self-weight of the exoskeleton on the human body increases, which prevents the knee joint from reaching the maximum flexion angle. Also starting from about 90% of the gait cycle, the plantar power output device begins to touch the ground which causes the flexion angle of the knee to no longer decrease. The curves of the angles of the ankle joint with wearing EF exoskeleton are also all above those when normally walking without wearing the exoskeleton, and the reason for this is the same as that for wearing NF exoskeleton.

In summary, during the walking of the human body wearing the exoskeleton, the exoskeleton has the greatest influence on the ankle joint comparing with other joints, which increases the plantar flexion angle. This is because the exoskeleton generates a driving force through the plantar power output device, and it will certainly cause certain influence on the ankle joint. The knee joint is also influenced, and with the increase in walking speed, the influence is increased. Although walking with wearing the exoskeleton influences the normal motion of the ankle and knee joints of the human body to a certain extent, it is still necessary to further verify whether the exoskeleton is unfavorable to the walking of the human body.

4.2. Dynamics

It can be seen from Fig. 12 that there are two large peaks in the torques generated at the ankle joint by the exoskeleton when walking at different speeds with wearing the NF exoskeleton that appear at

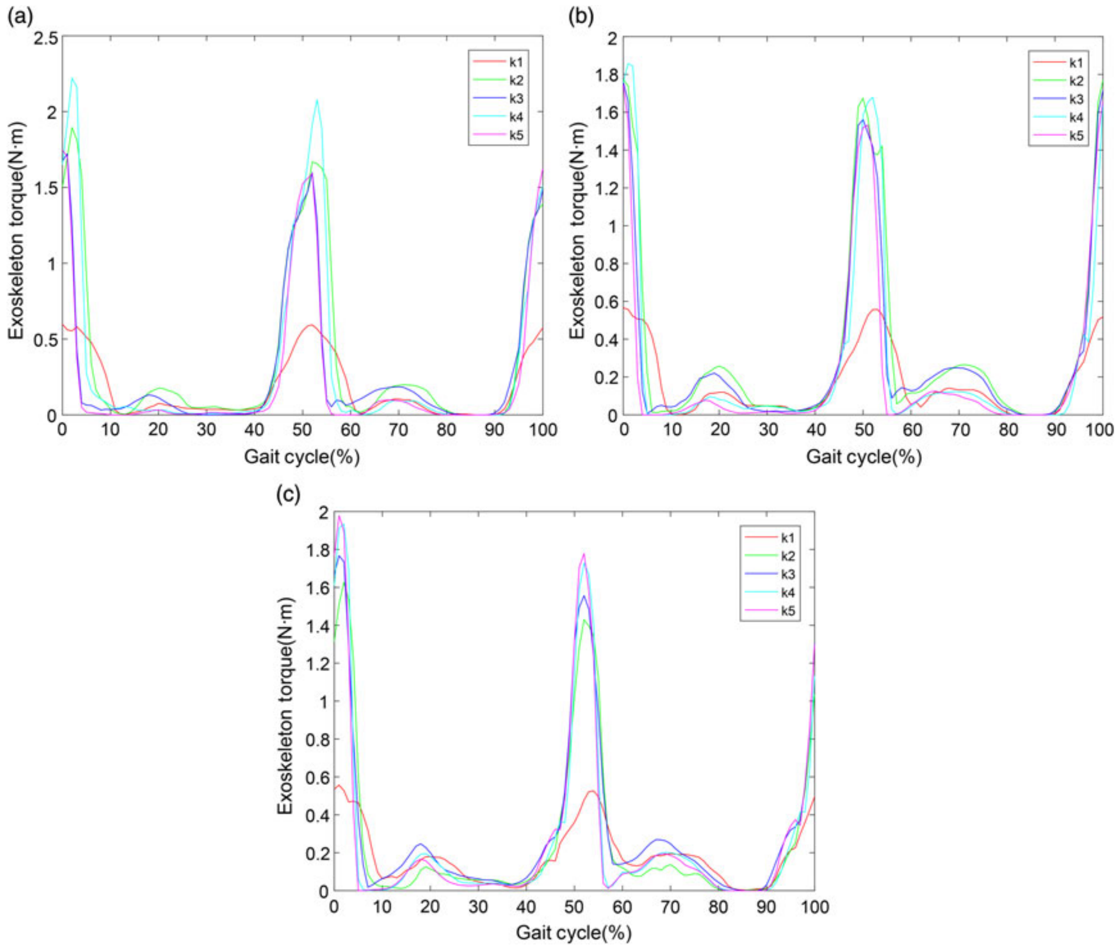

Fig. 12. Plantar flexion torque generated at the ankle joint by the exoskeleton: (a) walking at a speed of 0.6 m/s with wearing NF exoskeleton; (b) walking at a speed of 0.8 m/s with wearing NF exoskeleton; (c) walking at a speed of 1.0 m/s with wearing NF exoskeleton.

the beginning of the gait cycle and about 50% of the gait cycle, respectively. Taking the right foot, for example, when the right plantar power output device comes into contact with the ground, the exoskeleton generates a plantar flexion torque at the ankle joint of the left foot at this time and the right foot generates an almost equal plantar flexion torque simultaneously. At about 40% of the gait cycle, the left plantar power output device comes into contact with the ground and generates a plantar flexion torque at the ankle joint of the right foot. The plantar flexion torque peaks at about 50% of the gait cycle and decreases to 0 at about 57% of the gait cycle. It can be found by comparing the torque generated by the exoskeleton with the biological moment at the ankle joint that both have a peak plantar flexion torque at 50% of the gait cycle. However, it can be seen that the exoskeleton generates a very large plantar flexion torque at the beginning of the gait cycle which may probably cause certain influence on the ankle joint of the human body, but the general trend thereof is consistent with the biological moment of the human body and does not hinder the motion of the joint.

It can be seen from Fig. 13 that a flexion torque and an extension torque are generated at the knee joint and a plantar flexion torque is generated at the ankle joint by the exoskeleton when walking at different speeds with wearing EF exoskeleton. Since the force generated by the plantar power output device of the exoskeleton is provided to both the knee joint and the ankle joint in this state, the torques generated at the two joints influence each other. The diagram of the torque provided to the knee joint by the exoskeleton shows that the torque provided to the knee joint by the exoskeleton at each phase of the gait cycle is basically consistent with the biological moment and the trend of change is basically the same. The exoskeleton provides an extension torque to the knee joint at 0–30% of the gait cycle and a flexion torque to the knee joint at 40–63% of the gait cycle that reaches

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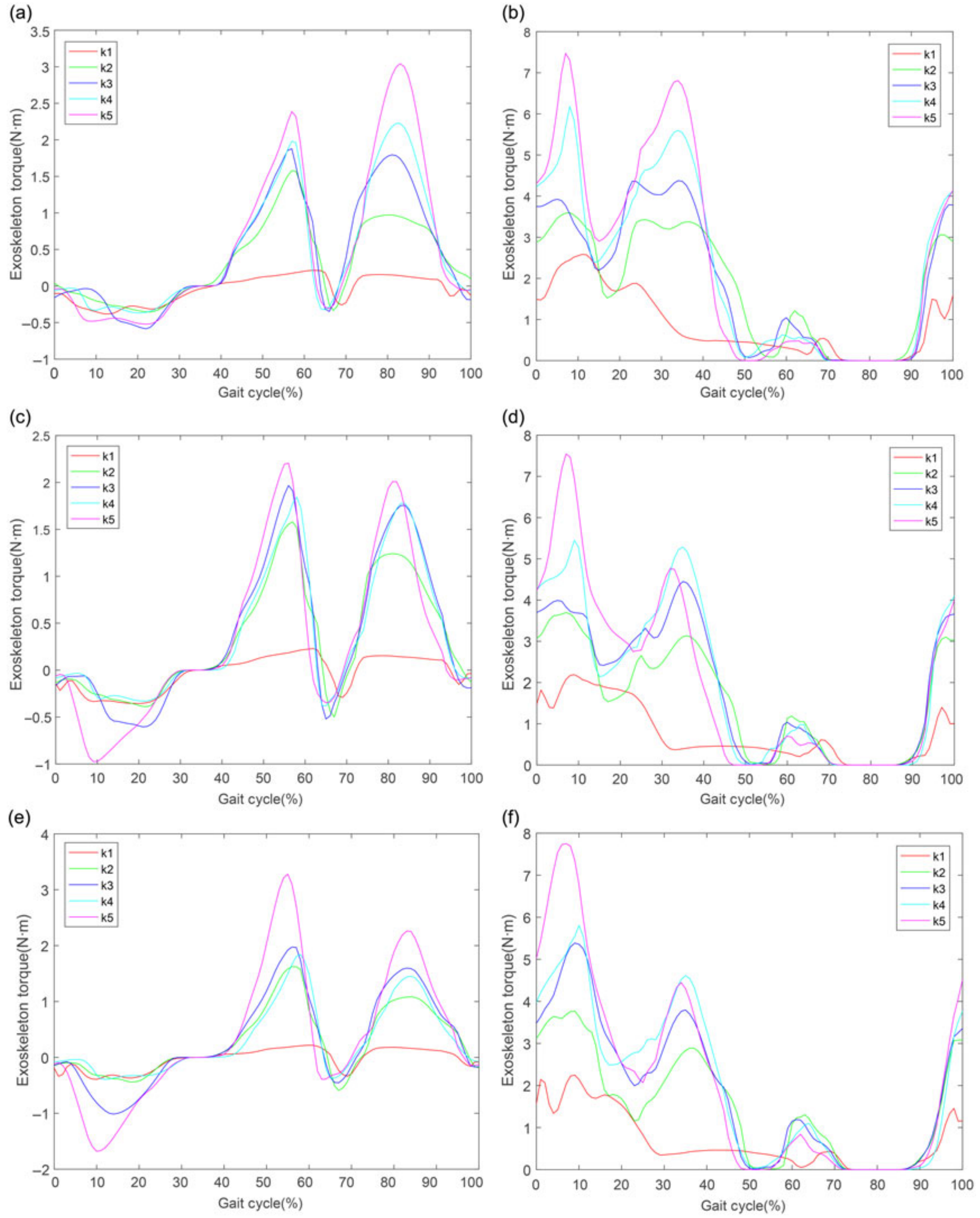

Fig. 13. Flexion torque (>0) generated at the knee joint by the exoskeleton and plantar flexion torque (>0) generated at the ankle joint by the exoskeleton: (a) torque generated at the knee joint when walking at a speed of 0.6 m/s with wearing EF exoskeleton; (b) torque generated at the ankle joint when walking at a speed of 0.6 m/s with wearing EF exoskeleton; (c) torque generated at the knee joint when walking at a speed of 0.8 m/s with wearing EF exoskeleton; (d) torque generated at the ankle joint when walking at a speed of 0.8 m/s with wearing EF exoskeleton; (e) torque generated at the knee joint when walking at a speed of 1.0 m/s with wearing EF exoskeleton; (f) torque generated at the ankle joint when walking at a speed of 1.0 m/s with wearing EF exoskeleton.

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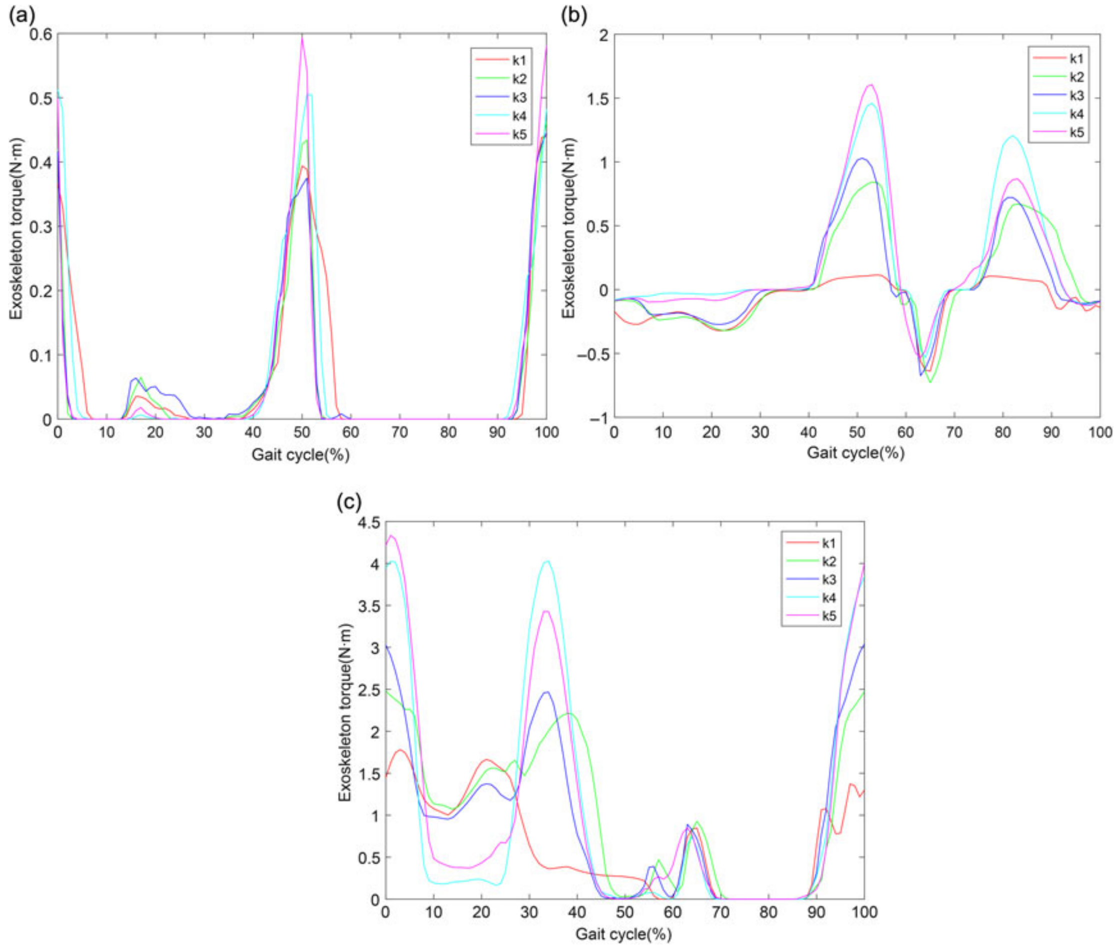

Fig. 14. Angle range of the knee execution unit providing assistive torque is 0°–40°, flexion torque (>0) generated at the knee joint by the exoskeleton and plantar flexion torque (>0) generated at the ankle joint by the exoskeleton: (a) torque generated at the ankle joint when walking at a speed of 0.6 m/s with wearing NF exoskeleton; (b) torque generated at the knee joint when walking at a speed of 0.6 m/s with wearing EF exoskeleton; (c) torque generated at the ankle joint when walking at a speed of 0.6 m/s with wearing EF exoskeleton.

a peak flexion torque at about 55% of the gait cycle. Starting from about 70% of the gait cycle, the knee joint is again provided with a flexion torque until the end of the gait cycle. The exoskeleton generates a plantar flexion torque at the ankle joint from the beginning of the gait cycle to about 50% of the gait cycle, which is the same as the phase of generating the plantar flexion moment by the human body. Although the growth trend is different, it provides the ankle joint with the plantar flexion torque required by the human body. It can be seen from the diagrams of torques that the greater the elastic coefficient of the spring used, the greater the torque generated at the joint by the exoskeleton. The smaller the elastic coefficient of the spring, the greater the fluctuation of the torque generated by the exoskeleton and the more unstable it is.

It can be seen from Fig. 14 that, after adjusting the angle range of the knee execution unit providing assistive torque as 0°–40°, the trend of the torque generated at each joint by the exoskeleton is basically the same as that when the angle range of the knee execution unit providing assistive torque is 0°–60°. However, due to the reduction of the assistance angle, the maximum elongation of the spring is reduced, and the torque as provided is reduced. By observing the diagram of the torque generated at the knee joint by EF exoskeleton, it is found that the torque range changes from 40–62% to about 40–56% and the duration of torque is shortened. The duration of the plantar flexion torque generated at the ankle joints is also shortened. Due to limited space, this paper only lists and analyzes the experimental results when the walking speed is 0.6 m/s.

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Table III. Average net metabolic cost of six subjects (mean ± standard deviation).

	Speed (m/s)					
	0.5	0.6	0.7	0.8	0.9	1.0
State	Net metabolic cost NEkg (W/kg)					
NW	1.777 ± 0.17	1.9128 ± 0.33	1.9558 ± 0.29	2.1461 ± 0.26	2.365 ± 0.32	2.5713 ± 0.42
k1EF	1.9704 ± 0.32	2.2235 ± 0.32	2.4611 ± 0.51	2.7574 ± 0.66	3.1286 ± 0.63	3.5262 ± 0.86
k2EF	1.9137 ± 0.18	2.0482 ± 0.29	2.2842 ± 0.36	2.586 ± 0.33	3.0075 ± 0.51	3.3742 ± 0.7
k3EF	1.5348 ± 0.26	1.7957 ± 0.3	2.1366 ± 0.52	2.5938 ± 0.47	2.9701 ± 0.55	3.384 ± 0.74
k4EF	1.7162 ± 0.33	1.899 ± 0.27	2.1434 ± 0.29	2.6844 ± 0.4	3.1021 ± 0.48	3.4505 ± 0.65
k5EF	1.6597 ± 0.22	1.9774 ± 0.25	2.2795 ± 0.34	2.4954 ± 0.31	2.8323 ± 0.52	3.3476 ± 0.63
k1NF	1.8974 ± 0.21	2.1932 ± 0.35	2.4018 ± 0.37	2.5903 ± 0.33	2.9762 ± 0.41	3.3005 ± 0.41
k2NF	1.7701 ± 0.34	1.995 ± 0.29	2.261 ± 0.31	2.628 ± 0.32	2.845 ± 0.31	3.2003 ± 0.48
k3NF	1.5731 ± 0.13	1.8088 ± 0.19	2.0071 ± 0.12	2.3314 ± 0.16	2.6322 ± 0.22	2.9343 ± 0.18
k4NF	1.7831 ± 0.3	2.0072 ± 0.28	2.3984 ± 0.35	2.6594 ± 0.33	2.9187 ± 0.33	3.2067 ± 0.44
k5NF	1.6676 ± 0.14	1.902 ± 0.22	2.1692 ± 0.25	2.3981 ± 0.26	2.7709 ± 0.28	3.1311 ± 0.33

k1EF and k1NF represent EF exoskeleton with k1 spring and NF exoskeleton with k1 spring, respectively, and so on.

By comparing the biological moments of the knee and ankle joints of the human body with the torques generated by the exoskeleton, it is found that the exoskeleton can basically provide required moment to the human body as appropriate and the hindrance to the human body is relatively small. By adjusting the angle range of the knee execution unit providing assistive torque, an adjustment of period which the assistance is applied to the knee joint of the human body can be preliminarily realized. The size of assistive torque can be adjusted by adjusting the elastic coefficient of the spring. Reasonably adjusting these two influence factors can make the exoskeleton better match the individual's gait.

4.3. Metabolic cost

The metabolic costs of the human body were tested at total of six walking speeds and on 10 different kinds of exoskeletons. Each subject needed to carry out 66 groups of experiments, and the whole experiment period was very long. Since the exoskeleton was passive, it was found that each subject's performance on walking with different exoskeletons varied greatly during the metabolic cost experiments. In order to get a more general rule, six subjects were analyzed statistically by SPSS software. Due to the long duration of the entire experiment, the state of the human body was different at each experiment. In order to reduce the influence of the fluctuations of the data acquired when standing for each group of experiment, a net metabolic cost NEkg was obtained by subtracting the data acquired during the first 2 min of standing from the data acquired during stable walking for each group of experiment when testing. The net metabolic cost of human walking with wearing different exoskeletons at different walking speeds is shown in Table III.

According to Table III, the net metabolic cost of NW increases with an increase in walking speed. When the human body walks with wearing exoskeleton, the net metabolic cost also increases with an increase in walking speed. In general, with the increase in walking speed, the metabolic cost of human walking with wearing exoskeleton increases from below-normal walking level to over-normal walking level. Then the exoskeleton changes from saving energy for human walking to increasing walking burden. No matter which speed the human body is walking at, the metabolic cost with wearing exoskeleton installed with k1 spring is almost the highest. From the analysis, k1 spring can produce very small force because its elastic coefficient is the smallest. So it cannot provide enough assistive force for human body. The exoskeleton begins to benefit the human walking only when the force provided by the spring was enough to overcome the burden of the exoskeleton's mass on the human body. The differences between NW and walking wearing exoskeleton were analyzed by paired t-test in SPSS software. Asterisk (*) indicates $p < 0.05$ and double asterisks (**) indicate $p < 0.01$. Figures 15 and 16 show the results of walking at a speed of 0.5 nd 0.6 m/s, respectively.

It can be seen from Figs. 15 and 16 that when walking at a speed of 0.5 and 0.6 m/s, respectively, the net metabolic cost of several EF exoskeletons and NF exoskeletons is lower than NW. This

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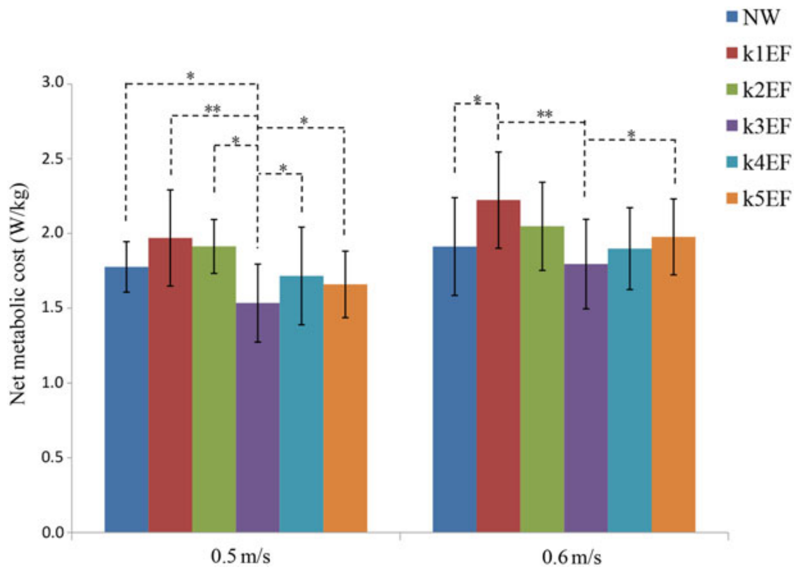

Fig. 15. Net metabolic cost when walking with wearing EF exoskeletons at a speed of 0.5 m/s and 0.6 m/s.

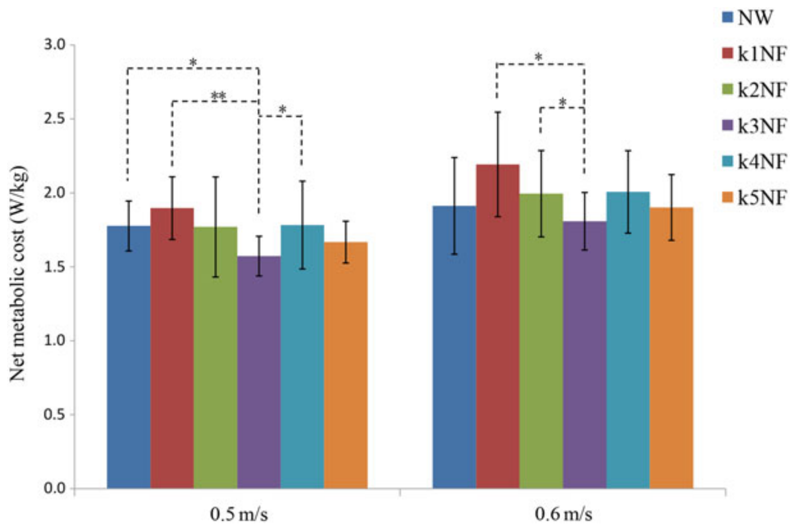

Fig. 16. Net metabolic cost when walking with wearing NF exoskeletons at a speed of 0.5 m/s and 0.6 m/s.

shows that some exoskeletons can save energy when walking at low speed, which is beneficial to walking. By comparison, when walking at a speed of 0.5 and 0.6 m/s, the net metabolic cost of wearing EF exoskeleton and NF exoskeleton installed with k3 spring is the lowest. This shows that the exoskeletons installed with k3 spring can save the most energy when walking at the two speeds compared with the exoskeletons installed with other springs. Walking with wearing the exoskeleton installed with k3 spring still consumes less than the exoskeletons installed with other kinds in most cases, even when the walking speed increases and the metabolic cost of walking with wearing the exoskeleton is higher than NW. Therefore, it can be considered that the k3 spring is the best of these five kinds of springs.

The net metabolic cost in the k3EF and k3NF states was subtracted from the net metabolic cost in NW state and then divided by the net metabolic cost in NW state to obtain the metabolic cost saving rate, as shown in Table IV. It can be seen that with the increase in the walking speed, the metabolic cost saving rate decreases. When walking with wearing k3EF exoskeleton at a speed of 0.5 m/s, the metabolic cost saving rate was the highest, reaching 13.63%.

In addition, it can be found that the net metabolic cost of the k3EF exoskeleton is lower than the k3NF exoskeleton at both speeds. This shows that the k3EF exoskeleton can save more energy than

Table IV. Metabolic cost saving rate.

Speed (m/s)	0.5		0.6	
State	**NEkg (W/kg)**	**Saving rate (%)**	**NEkg (W/kg)**	**Saving rate (%)**
NW	1.777		1.9128	
k3EF	1.5348	13.63	1.7957	6.12
k3NF	1.5731	11.47	1.8088	5.44

k3NF exoskeleton for human walking. Because the k3NF exoskeleton, when being worn for walking, can provide an assistive torque to the ankle joint, it can thus reduce the metabolic cost of the human body when walking. When walking with wearing the k3EF exoskeleton, since the exoskeleton can provide not only an assistive torque to the knee joint but also an assistive torque to the ankle joint, it can thus reduce more metabolic cost for the human body. However, when walking speed increased to 0.7 m/s, the net metabolic cost of walking with wearing exoskeleton began to be higher than normal walking. We think that there are some reasons for this result. First, the mass of the exoskeleton especially the plantar power output device is a little heavy, which changes the inertia of the legs and increases the metabolic cost of acceleration and deceleration. With the increase in walking speed, the influence of the inertial force generated by the self-weight of the exoskeleton on the human body also increases. Second, when the walking speed increases, the straps loosen and the exoskeleton may slide down, thus joints between the exoskeleton and the knee dislocate, which hinders human walking.

In summary, when the human body walks at a speed below 0.7 m/s, the exoskeleton can save metabolic cost. When walking with wearing the exoskeleton installed with k3 spring, it will bring the most benefits to the human walking and produce the least metabolic cost compared with other exoskeletons. When walking with wearing k3EF exoskeleton at a speed of 0.5 m/s, it can save 13.63% of the metabolic cost for the human body. For the elderly and the people with knee injuries, because of the loss of muscle strength and joint injury, their mobility is decreased and their walking speed is relatively low. The exoskeleton can save energy consumption for these people and reduce walking burden effectively.

5. Conclusions

We have developed a passive lower extremity exoskeleton with a simple structure and gait self-adaptivity. The exoskeleton takes the human body's gravity as a driving force and makes a crank-slider mechanism to perform a reciprocating motion through pulling forces of Bowden cables so as to finally generate torques at ankle and knee joints. The exoskeleton provides a flexion torque to the knee joint during the terminal stance phase, the pre-swing phase, the mid-swing phase and the late-swing phase and provides a plantar flexion torque to the ankle joint during stance phase. The exoskeleton also provides a torque for assisting the extension of the knee joint during midstance phase to make the knee joint stable and reduce the risk of falling due to insufficient leg support force. The design idea of the exoskeleton has opened up new ideas for the design of passive exoskeletons. At present, the exoskeleton can only provide assistance to the human body when walking horizontally. We designed a locking device in order to facilitate the use in daily life for people. When the locking device is released, the exoskeleton no longer interferes with the normal motions of the knee joints of the human body. It has been found through experiments that wearing the exoskeleton has certain influence on the motion of the ankle and knee joints. The assistance function of the exoskeleton was verified through dynamic experiments, and it is found that the size of the torque and the period of providing the torque can be adjusted by changing the elastic coefficient of the spring and the angle range of the knee execution unit providing assistive torque. Finally, the metabolic cost of the human body when walking with wearing different exoskeletons at various speeds was tested. The results show that the k3EF exoskeleton saved the most metabolic cost for the human body when walking at a speed of 0.5 m/s and the saving rate is 13.63%. With the increase in the walking speed, the metabolic cost saved for the human body by the exoskeleton decreases. We will further optimize the structure of exoskeleton to reduce the mass of the exoskeleton, especially optimizing the plantar power output device. In the future, the length of the link can be adjusted by designing a link adjustment mechanism so as to make the exoskeleton assist the human body to sit-to-stand, climbing stairs and slope and so on.

Acknowledgments

This study received financial support from Basic Research Foundation of Beijing Institute of Technology, China (20160042030) and Research Foundation for Soft Exoskeleton R&D, Beijing Advanced Innovation Center for Intelligent Robots and Systems, China (1780037231707).

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