Improving the Assisting Efficiency of Ankle Robot through Energy Harvesting of Achilles Tendon

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Abstract

Background: The construction of lightweight robots poses one of the major challenges in the field of active robots since bearing the weight of an active robot significantly increases metabolic cost. However, few studies have achieved a substantial reduction in the robot weight. The primary reason is that the weight of the actuator, which comprises the main weight of the robot, is limited by the specific power, power requirements and assisting efficiency.

Methods: In this paper, we propose a new method that is utilizing the energy harvesting function of the Achilles tendon to improve the assistance efficiency of ankle robots to reduce the weight of the actuator and we design a novel ankle robot to test the validity of the method. The robot works with the ankle plantar flexor at 43% -60% of the gait cycle and has no other effects on the joints or the tendon of the lower limb. Healthy subjects were recruited to test the prototype in three conditions: free walking, power-on walking, and power-off walking. Data on the robot assisting power, metabolic cost and kinematics in different conditions were collected and analyzed.

Result: The results showed that the ankle robot can deliver forces at the controlled assistance timing. The average assisting power of 0.0650±0.0054 W/kg per leg resulted in an 8.7±8.1% and 19.0±6.4% net reduction in metabolic cost in power-on walking compared to free-walking and power-off walking, respectively.

Conclusion: Compared with some of the best research, our initial result supports the validity of the method. This method can help to reduce the weight of active robots and the technical innovative method to determine the assistance timing more accurately and the novel design of the ankle robot can provide a reference for future research. To the best of our knowledge, this method is the first to use the human physiological structure to optimize the design of active robots.

Keywords: Assisting efficiency, Ankle robot, Energy harvesting, Achilles tendon, assistance timing

Background

Humans take more than 10,000 steps per day on average [1], and the metabolic cost of walking is higher than those of other daily activities [2]. Therefore, reducing metabolic cost is of great significance to the preservation of physical strength.

To reduce the metabolic cost of walking, much research has been performed regarding designing and using wearable lower-limb robots to assist the motion of the joints [3]. In general, wearable lower-limb robots can be concluded as devices that use structural parts to connect with the segments of the lower limbs of humans and use a mechanism to actuate the relative motion between the structural parts. According to whether there is
an external power source or not, the lower limb robots can be divided into active robots and passive robots [4]. The principle of passive robots is using elastic elements between the segments of the lower limb to imitate or replace the tendon to absorb the negative work and assist the positive work of joints during walking. However, the stiffness of the elastic element needs to be designed strictly according to the subject and motion conditions [5], which limits the practical use of passive robots. Active robots use external controllable power sources such as electricity or pneumatics to actuate robots, which makes active robots adaptable to various requirements in practical applications.

A variety of active robots have been developed in the past two decades. Early robots were designed to assist the multiple joints of the lower limbs in walking [6-8]. These robots had a typical rigid structure in which the segments of the robot align with those of the lower limb and the joints of robots are coaxial with those of the lower limb. However, the heavy weight of the mechanical structure greatly increased the metabolic cost [7] and the inconsistency of the human-machine structure affects the movement of humans and function of the tendon [3].

Some of the later studies fixed the robots on a frame to reduce the burden on humans, which reduced metabolic cost. Philippe and his colleagues designed a robot that assisted the movement of the ankle joint by pneumatic muscles in 2013 and realized 6±2% reduction in the metabolic cost [9]. In 2017, Galle and his colleagues designed a robot with a similar structure. They tested different timings of assistance and trajectories of assisting power and found that assisting the ankle joint at 42% to 60% of the gait with 0.4 W/kg assisting power can achieve an optimal reduction in metabolic cost(21%) [10]. These robots still had the problems of the inconsistency of human-machine structure. In the same year, B.Quinlivan and his colleagues tested a multi-joint soft exosuit and achieved $22.83 \pm 3.17\%$ net metabolic cost reduction with an assisting torque of 38% biological ankle moment [11]. The robot was joint-less and did not restrict the joint movement structurally, but the results showed that the dorsiflexion of the ankle decreased greatly, which affected the function of the Achilles tendon. Although these robots reduced the net metabolic cost, most of these robots are only suitable for applications on a treadmill such as the walking training of patients and cannot be used in actual moving conditions [12, 13], since they are fixed to a frame.

Recently, some lightweight autonomous robots have been designed to assist joints during walking. In 2014, Alan and his colleagues designed a multi-joint soft exosuit to assist the movements of lower limbs, but the results showed an increase in the metabolic cost [14]. An important reason was that the excessive weight of the robot increased the metabolic cost, and there were also some possible reasons that the fabric was so flexible that it consumed energy from the motors [15], and the negative power of the robot affected the ankle dorsiflexion [16]. In 2015, Luke and his colleagues developed a novel joint-less robot to assist the plantarflexion of the ankle joint [4]. Although the robot achieved $11\pm4\%$ net metabolic cost reduction with the assisting power of $0.105\pm0.008$ W/kg per leg, the weight of the actuators (approximately 1.1kg/leg) borne by the shank led to a metabolic cost higher than that borne by the back [17]. Ye Ding and his colleagues designed a robot to assist hip extension, which achieved a reduction in net metabolic cost of $8.5\pm0.9\%$ in 2016 [18] and achieved a $9.3 \pm 2.2\%$ reduction in net metabolic cost in another study in 2019 [19].
Improvement in the robots is mainly reflected in improvement in the assisting efficiency, including 1) only assisting the movement of the joints that require high power; 2) eliminating the restrictions of the robot on human joints and not affecting the natural movement of the lower limbs; 3) selecting suitable assistance timing; and 4) reasonably distributing the load. The assisting efficiency refers to the ratio of net metabolic cost reduction in the assisting power. Although these methods have played certain roles, the key problem affecting the efficiency of assistance has not been solved, that is the excessive weight of the active robot. Take the example of the robot designed by Alan and his colleagues in 2014. The design of the robot utilized all the efficient assistance methods above, but the results showed that the net metabolic cost increased by 9.3% [14]. An important reason for this result was that the weight of the robot (10.1 kg) increased the metabolic cost by 16%-17.5%. In the composition of an active robot, the actuator occupies most of the weight of the robot. Specifically, in a robot driven by motors, the weight of motors and reducers always constitutes a large proportion of the weight. However, simply reducing the weight of the motor and reducer causes the robots to fail to meet the power requirements because the weight of the motor and the reducer is related to the specific power.

Considering that the Achilles tendon can store and return 61% of the energy of the ankle joint [20] and that this function has been ignored or affected in previous studies, in this paper, we propose a new method to improve the assisting efficiency of active robots through the energy harvesting of the Achilles tendon during walking with a robot to assist ankle plantar flexion. Different from the methods above that improve the assisting efficiency by reducing the metabolic cost, our method can reduce the assisting power of the robot, which can reduce the weight of the motor and reducer.

To test the validity of the method, we designed a novel active robot to assist the plantarflexion of the ankle joint. The prototype can retain the role of the tendon to harvest energy and provide positive work assistance at the right time. The prototype has a joint-less structure, an optimal load distribution and lighter drivers, and we improve the wearing stability, the rigidity of human-computer interaction and the wearability by designing a 3D printing thermoplastic polyurethanes (TPU) calf connector. We propose a new method to determine the assistance timing more accurately based on a commercial gait analysis system. To balance the force exerted on the calf and the assisting force, we used the current-based semi-closed loop control. The control method also provided security in the human-computer interaction. Five healthy subjects were recruited in the experiment to test the effect of the robot on human walking and the results showed that 0.0650±0.0054 W/kg per leg assisting power resulted in an 8.7±8.1% reduction in metabolic cost with insignificant gait changes. Compared with some of the best research, the initial results support the validity of the efficient method.

The greatest innovation and contribution of this paper is the proposal of an efficient assisting method. Moreover, the novel mechanism design, hardware selection, innovative method of assistance timing determination and control method in this paper can provide a reference for future research.

**Method**

**Prototype design**

The prototype is designed to assist the ankle plantarflexion which takes 43% of the positive
work [21] and the main forward driving force [16]. The prototype can be divided into three parts by function: controlling part, driving part and actuating part (Fig.1). The actuators with smaller power are selected to reduce the weight of the robot because of the proposed efficient assisting method. (The weight comparison is shown in Table S1 in an additional file.) For the optimal weight distribution, the controlling part and the driving part that comprise most of the robot weight are borne on the back and the actuating part has a lightweight structure. The Bowden cable is used to transmit the power, which eliminates the limitation on the degree of freedom of the ankle joint for the prototype. In addition to following the efficient-assisting design above, the design of the prototype realizes the separation of the drivers and actuators, which can reduce the effect on the movement of other joints. In addition, the improved rigidity of the actuating part and the transmission of the Bowden cable make the prototype have both the flexibility of a soft robot and the response performance of a rigid robot. Ergonomics is also applied to the design of the prototype.

Specifically, the controlling part of the prototype comprises an IPC (Germany, Beckhoff, C6015-0010), and two controllers (Switzerland, Maxon, EPOS4 COMPACT 50/15 EtherCAT) (Fig.2 A). The controlling program is written in IPC (Germany, Beckhoff, C6015-0010) and commands the controllers (Switzerland, Maxon, EPOS4 COMPACT 50/15 EtherCAT) to control the movement of the motors through the EtherCAT bus.

The actuating part comprises two motors (Switzerland, Maxon, EC 60 flat, 150 W, 647694), two gear reducers (Switzerland, Maxon, GP 52 C, 43:1, 223089), a wire wheel, a wire box, and a Bowden cable (Fig.2 B). The maximum speed of the motors, the ratio of the reducers and the size of the wire wheel are selected according to walking biomechanics. The motor (Switzerland, Maxon, EC 60 flat, 150 W, 647694) connects with the reducer (Switzerland, Maxon, GP 52 C, 43:1, 223089) and the wire wheel in series, which can provide a maximum continuous force of 366 N and attain a maximum speed of 0.254 m/s. Then the wire box and the Bowden cable transmit the power. For less friction of the Bowden cable, a lower wrapping angle is required [22]. Thus, the interface of the Bowden cable on the wire box is designed to face down along the trunk and the interface of the Bowden cable on the calf connector is hinged to the calf connector to maintain a lower wrapping angle. The hinge-wheel guiding mechanism and idler wheel set on the calf connector are used to guide the direction of the inner cable of the Bowden cable, which guarantees the direction of the tension force in the direction of the tibia (Fig.2 C). When the tension force is generated, the calf bears the force and transmits it to the ground in the stance phase, so it has no effect on the hip.
joint and knee joint.

The driving part is lightweight for comfort, comprising two plastic calf connectors and a pair of lightweight running shoes. The calf connector is made of three layers of different materials (Fig. 2 C). The inner layer is a cushion made of porous sponge for comfortable wearing and has a large coefficient of friction. The middle layer is made of TPU material through 3D printing. The flexibility of the TPU in the bending direction can make the calf connector bend to adapt to the shape of the calf, while the rigidity of the material in the stretching direction gives the calf connector a high interaction stiffness. The outer layer is a locking mechanism for fixing the calf connector to the calf. For shoes, a 3D printing anchor made of ABS is fixed on the sole of ordinary sports shoes. The size of the anchor is designed to guarantee an appropriate ankle force arm and avoid interference of the Bowden cable with the calf muscle.

Two kinds of force sensors are mounted between the Bowden cable and anchor to sense the interaction between the human and the prototype. One is the load cell (Germany, WIKA, Tecsis, F2811) that measures the tension between the inner cable and the heel anchor. The load cell’s amplifier (Germany, WIKA, Tecsis, B1936) is connected with the analog input terminal (Germany, Beckhoff, EL3124) which sends the data of the load cells (Germany, WIKA, Tecsis, F2811) to the IPC (Germany, Beckhoff, C6015-0010) through the EtherCAT bus. The other one is a gait analysis system (Germany, H/P/cosmos, Mercury, FDM-THM, 120 Hz) that transmits the collected plantar pressure data to the IPC (Germany, Beckhoff, C6015-0010) through USB2.0. The IPC (Germany, Beckhoff,
C6015-0010) determines the assistance timing by the plantar pressure and gait cycle.

The complete route of the data stream and power flow is shown in Fig.3. The IPC (Germany, Beckhoff, C6015-0010) determines the assistance timing from the plantar pressure collected by the gait analysis system (Germany, H/P/cosmos, Mercury, FDM-THM, 120 Hz) and then sends out the control command to the controller (Switzerland, Maxon, EPOS4 COMPACT 50/15 EtherCAT) to control the movement of the actuator. The force generated from the actuator makes the Bowden cable pull the ankle joint and the force is detected by the load cell (Germany, WIKA, Tecsis, F2811). The IPC records the stretching force from the terminal (Germany, Beckhoff, C6015-0010).

Ergonomics is considered throughout the design of the prototype. The weight center of the backload is placed at the approximate level of thoracic vertebrae 1-6, which is the optimal load-bearing position regarding energy [23]. The weight of the backload is limited to less than 10% of the person’s bodyweight to reduce the impact of the weight of the backload on kinematics and avoid the appearance of subjective weight-bearing sense and back pain [24]. A waist belt is used to improve the load distribution [25], to reduce the load feeling and improve the perceived stability during walking [26]. To increase the contact surface and reduce the pressure [27], the shape of the calf connector is similar to that of the gastrocnemius muscle.

Fig.3 The complete route of the data stream and power flow of the prototype. Each block represents a function component of the prototype, and the function and component of the function component are shown in the block. The red arrow indicates the direction of the data stream and power flow that run out from the prototype, while the blue arrow indicates the direction of the data stream that returns to the prototype.
Assistance timing

As shown in the Fig.4, the ankle joint performs the main negative work in the first 43% of the gait cycle and the main positive work in the 43% to 60% of the gait cycle. In the first 43% of the gait cycle, the Achilles tendon stores most of the negative work, and in 43% to 60% of the gait cycle, the Achilles tendon returns energy to help the plantar flexor achieve plantarflexion of the ankle. Therefore, the timing of assistance is set in 43% to 60% of the gait cycle to utilize the energy harvesting function of the Achilles tendon, and at other times of the gait cycle, the prototype does not exert any influence on human movement. The timing of assistance was tested and considered the optimal assistance timing in the study [10].

Foot switches [10] or inertial sensors [28] are usually used to judge the gait event which is used to determine the assistance timing with the gait cycle in the traditional method. The foot switch at the bottom of the heel detects the gait event when the heel strikes the ground, but the position of the foot switch and the individual gait difference of people often lead to errors in the result. The inertial sensor detects the gait event when the joint reaches a specific position such as the extreme position, but the muscle movement and the judgment algorithms often result in errors. Therefore, we propose a more accurate method to determine the assistance timing with the help of a commercial gait analysis system.

In this method, the assistance timing is judged by the ground reaction pressure which is detected by the gait analysis system (Germany, H/P/cosmos, Mercury, FDM-THM,120 Hz) and is read out by Zebris hardware SDK (Germany, Zebris Medical GmbH) to a computer that communicates with the IPC (Germany, Beckhoff, C6015-0010) through USB 2.0. A quick method of image processing is used to determine the assistance timing in real time (<5 ms) [29]. First, the pressure data are transformed into a grayscale image by OpenCV 3, as shown in Fig.5 A. Second, a 9×5 mask is used to dilate the Fig to obtain a connected area, as shown in Fig.5 B. Third, we obtain the edge through the Sobel gradient operator, as shown in Fig.5 C. Fourth, we calculate the pressure center within the area wrapped by the edge, as shown in Fig.5 D. The equation calculating the pressure center is as follows:

Different subjects share similar gait characteristics. A ankle power, where the red area represents positive work and the blue area represents negative work. B ankle angle corresponding to the movement.

Fig.4 Power and angle of the ankle joint during walking. The data were collected from a healthy man who walked at 1.05 m/s (self-selected speed).
\[
\begin{align*}
X_{COP} &= \frac{\sum F_i x_i}{\sum F_i} \\
Y_{COP} &= \frac{\sum F_i y_i}{\sum F_i}
\end{align*}
\]

where $X_{COP}$ and $Y_{COP}$ are the coordinates of the pressure center; $x_i$ and $y_i$ are the coordinates of the pixel within the area wrapped by the edge; and $F_i$ is the pixel value of the corresponding pixel. The origin of the coordinate is at the lower-left corner, the rightward direction is the positive direction of the $x$-axis, and the upward direction is the positive direction of the $y$-axis. Fifth, if the abscissa $X_{COP}$ is less than 28 which is the abscissa of the middle line, the left foot is considered to be touching the ground; otherwise, the right foot is. Sixth, once the gait analysis system detects pressure, the computer uses the above method to determine which foot is touching the ground, and the moment is set as the starting point of the gait cycle of the lower limb of the corresponding side. The start time of the prototype assistance can be calculated by the equation: $t_{\text{start}} = 0.4 \times t_{g, \text{cycle}}$ and the end time can be calculated by the equation: $t_{\text{end}} = 0.6 \times t_{g, \text{cycle}}$ where $t_{\text{start}}$ is the start time of the prototype assistance; $t_{\text{end}}$ is the end time; $t_{g, \text{cycle}}$ is the gait cycle which is the time interval between the same-side foot striking on the ground twice in the previous gait cycle in free walking. The $t_{g, \text{cycle}}$ of different subjects are collected before the experiment.

**Fig.5** Process of determining which foot is touching the ground. A result of the ground reaction pressure converted to an image. The 56×128 pixels represent the 56×128 pressure detecting unit in the gait analysis system (Germany, H/P/cosmos, Mercury, FDM-THM, 120 Hz) and the whiteness represents the pressure. To clearly show, the value of pressure is magnified ten-fold. B result of two dilatations through a 9×5 mask. C result of contour edge extraction. D result of adding C to A.

**Control method**

Studies have shown that greater assisting force leads to lower metabolic cost [11]. However, excessive tension can cause excessive pressure on the calf, affecting the wearing comfort and blood flow [27]. To balance these two situations, we control the output torque of the motor according to the subjective feelings of the subjects. Moreover, due to the short boost time in a single cycle, there is a high demand for a control response. Therefore, a half closed-loop current control is designed to control the torque of motors and obtain a faster control response. A PID
controller is designed for the controllers (Switzerland, Maxon, EPOS4 COMPACT 50/15 EtherCAT) to track the set current value, and the parameters of the PID controller are tuned manually to achieve a quick response (Fig.6). According to the characteristics of the DC servo motor, the torque of the motor is proportional to the current. The relationship between the torque and the current can be written by the formula: 

\[ T_m = K \cdot I \]

where \( T_m \) is the torque of the motor; \( K \) is the torque coefficient of the motor and \( I \) is the current of the motor. A small initial current value is set and then increases slowly according to the requirements of the subjects. During the robot assistance, the motor output torque is controlled by the above method. After that, the motor quickly resets to the initial position and waits for the next assistance timing.

\[ \begin{align*}
&\text{Current setting} \quad I_s \\
&\text{PID controller} \quad I_{\text{com}} \\
&\text{Robot} \quad T_f
\end{align*} \]

Fig.6 Diagram of the actuation scheme. During the assisting time, the PID controller outputs the controlled current to the controllers to boost the robot (Switzerland, Maxon, EPOS4 COMPACT 50/15 EtherCAT) based on the set current \( I_s \) and the feedback current \( I_f \). \( T_f \) is the interaction force between the robot and human.

Safety guarantee
To ensure the safety of the experiment, five protection measures are taken. 1) Position limit by a spring with high stiffness. The position of the stiff spring limit is set to the approximate limit position of the plantar flexion before the experiment. The flexibility of the spring protects the motor from the impact of the emergency stop. 2) Anti-fall rope. The anti-fall rope is used to prevent falling caused by accidents. 3) Human-machine interaction peak force limit. When the human-machine interaction force exceeds the peak force limit, the prototype is powered off. 4) Emergency switch. The prototype is equipped with an emergency switch. When the subject feels any discomfort, the emergency switch can be pressed to power off the prototype. 5) Detection of abnormal gait. When the gait analysis system (Germany, H/P/cosmos, Mercury, FDM-THM, 120 Hz) detects that the foot on the same side has struck the ground twice consecutively, the prototype does not work in this gait cycle until the system detects the feet on both sides striking the ground alternatively.

Participants
Five subjects (age: 23.8±0.9 years old, height: 167.9±1.8 cm, and weight: 58.6±2.3 kg) were invited to participate [30]. The subjects were confirmed to have no history of neurological diseases or injuries, joint diseases, musculoskeletal diseases or injuries, visual or proprioceptive disorders, or metabolic diseases. All the subjects were informed of the purpose and process of the experiment and signed a written consent form.

Experimental protocol
A total of three conditions were used for the evaluation: free walking, power-off walking, and power-on walking. The subjects were asked to walk on a gait analysis system (Germany, H / P / cosmos, Mercury,
FDM-THM, 120 Hz) for 6 minutes. The walking speed was 1.0 m/s, which was the average speed of the subjects in the pre-experiment where the subjects were asked to walk on the ground at a self-selected speed. For each subject, all three conditions needed to be tested, and the order was random. Before the start of each test, the subjects rested for 10 minutes. In the first minute of power-on walking, the assisting force was adjusted according to oral feedback regarding the subject’s subjective feeling until they felt the maximum assistance while comfortably wearing the prototype. One day before the formal experiment, each subject performed three groups of 6-minute power-on walking to accommodate the assistance of the prototype. The experimental method is shown in Fig. 7.

Fig. 7 Experiment methods. Data was collected from a subject walking on the H/P/ COSMOS gait analysis system in different conditions. Metabolic cost was measured by means of potable gas analysis system (K5b², Cosmed, Roma, Italy) and the subject’s kinematics are measured by means of a three-dimensional gait analysis system (Vicon, Oxford Metrics, UK; 250 Hz).

Mechanical power
The mechanical work performed by the prototype to the human can be calculated by the equation: \[ P = \sum \frac{F \times \Delta S}{\Delta t} \], where \( F \) is the assisting force, which is detected by the load cell (Germany, WIKA, Tecsis, F2811) and smoothed by an average window function whose size is five command cycles (5 ms); \( \Delta S \) is the length change of the Bowden cable in a collecting cycle of the assisting force; and \( \Delta t \) is the time of assisting the movement of the ankle. \( \Delta S \) can be calculated by the formula: \[ \Delta S = \frac{2 \pi r \Delta \theta}{K_e} \] where \( r \) is the radius of the wire wheel; \( \Delta \theta \) is the change in the output of the
encoder mounted on the motor; $K_c$ is the number of subdivided pulses per revolution; and $i$ is the reduction ratio of the reducer.

**Metabolic cost**
The metabolic cost was measured by a portable gas analysis system (K5b®, Cosmed, Roma, Italy). Carbon dioxide and oxygen were used to calculate metabolic cost through Brockway's standard formula [31]. The average of relatively stable values over one minute was used to represent the metabolic cost. The metabolic cost of standing was determined for each subject before the experiment. The net metabolic cost was obtained by subtracting the standing metabolic cost from the walking metabolic cost of each condition and was normalized by the weight of the subjects.

**Kinematics**
A three-dimensional gait analysis system (Vicon, Oxford Metrics, UK, 250 Hz) was used to analyze the kinematics. Sixteen markers were stuck to the lower limb according to the requirements of the lower limb plug-in-gait model [32]. Sagittal kinematics were attained by using multiple rigid body kinematics (Nexus, Oxford Metrics, UK). There were some differences in the marker position on the posterior superior iliac spine among different conditions since the prototype affected the marker position. This caused different initial joint angles for the hip joint and knee joint. Therefore, we analyzed the range of the joint motion. The average of the gait parameters of 20 stable steps were taken as the gait parameters of 1-step.

**Statistical methods**
SPSS (SPSS Inc., USA) was used to conduct the statistical analysis [33]. A paired-samples t test with the three conditions was used to verify the effect of the device on human walking. $p < 0.05$ was set as the level of a significant difference.

**Result**

**Mechanical power**
The average peak interaction force was 126 ± 10 N. The maximum peak interaction was 162 N. The average mechanical power was 0.0650 ± 0.0054 W/kg. Fig.8 shows the trajectory of the interaction force between robot and human in a gait cycle corresponding to time and the condition of the Bowden cable. The detailed mechanical power is presented in the additional file (Table S2).

![Fig.8](image)

**Fig.8** Human–robot interaction. The trajectory of the interaction force between robot and human corresponding to time and the condition of the Bowden cable during a gait cycle is shown.

**Metabolic cost**
The average metabolic power during standing was 1.9 ± 0.1 W/kg. The average net metabolic power was 4.3 ± 0.4 W/kg, 4.7 ± 0.5 W/kg, 4.2 ± 0.5 W/kg during power-on walking, power-off walking, and free walking, respectively (Fig.9). The net metabolic cost in the power-on walking was 8.7 ± 8.1% ($p=0.427$) and 19.0 ± 6.4% ($p=0.044$) lower than that during the free walking and power-off walking, respectively. The net metabolic cost during the power-off walking was 12.6 ± 3.6% ($p=0.034$) lower than that during free walking.
Each subject’s metabolic cost in the three testing conditions is presented in the additional file (Table S3).

Fig. 9 Metabolic power in the three test conditions: free walking (green), power-on walking (yellow), and power-off walking (purple). Data are shown as mean±SD.

Kinematics

Fig. 10 shows the range of the joint motion for the subject who achieved the highest reduction in the metabolic cost during power-on walking compared to that during free walking. There was almost no difference in the ranges of the hip angle and knee motion between the conditions, but there was a difference in that of the ankle joint. In power-on walking, the subject had less plantarflexion, which exactly showed the effect of robot assistance on the plantarflexion of the ankle joint.

Discussion

The initial result supported the idea that utilizing the energy harvesting function of Achilles tendon can improve the assistance efficiency of ankle robots. The average interaction power of 0.0650±0.0054 W/kg per leg caused an 8.7 ± 8.1% reduction in net metabolic cost during power-on walking compared to that of free walking, which is more efficient than the results of the study where 0.105±0.008 W/kg per leg caused an 11 ± 4% reduction [4]. A rough calculation shows that for every 1% reduction in metabolic cost, the interaction power in this study is 21.7% lower than that in the referenced study, where the initial preload affects the ankle dorsiflexion and the energy storage of the Achilles tendon [4]. The average peak interaction force of 116.3 N in our study is approximately one-fourth of that in study [11] and one-third of that in study [33], but we obtained a similar reduction in metabolic cost. If the reduction in the metabolic cost in power-on walking compared to that in power-off walking is regarded as the net effect of robot assistance, the 19.0 ± 6.4% reduction in the metabolic cost caused by the 0.0650±0.0054 W/kg per leg assistance in our study is 31% more efficient than the results of the study where 0.4W/kg assisting power cause 21% reduction in metabolic cost [10].

In terms of the amount of the net reduction in metabolic cost, our robot achieved a result similar to those of some of the best research at present. The average net metabolic cost reduction of 8.7±8.1% in power-on walking compared to free walking was slightly more than the 7.3±5.0% in study [33] and slightly
less than the 11±4% in study [4] and the 9.3% in the study that designed a hip robot [19]. The average net metabolic cost reduction of 19.0±6.3% in power-on walking compared to power-off walking is slightly less than that of 21% in study [10] and 22.84±3.17% in study [11].

It should be noted that the effect of the assistance of the active robot on the metabolic cost during walking seems to vary from person to person. In this study, there was a "super subject", who showed a large net reduction in metabolic cost with robot assistance compared to all the other subjects. There are "super subjects" in other studies [4, 19], although the variance in other studies is not as large as that in this study. In a sense, the results of the super subjects can represent the optimal results of the robot's impact on humans. From the perspective of this point, we achieved the largest reduction in the net metabolic cost, since the "super subject" showed the highest known net metabolic cost reduction of 32.5% in power-on walking compared to that in free walking in this study. To verify the results, we repeated the experiment on this subject and obtained similar results. These results also supported the proposed method. There was an increase in metabolic cost of two subjects in power-on walking compared to in free walking. They appeared to be more sensitive to the backload during walking, and less sensitive to the assistance. Other possible factors, such as the position of the backload, the comfort of wearing, the timing of the power-up, and the individual habit of exerting the strength of the ankle muscles may affect the metabolic cost. Due to the differences in the net metabolic cost reduction from person to person, selecting different subjects seems to have different effects on the results. Therefore, a further study should be conducted with many more subjects to test the method and determine the cause of the variation from person to person in the reduction in metabolic cost.

The reaction force of the interaction force is borne by the calf, so the peak interaction is limited by the pressure that the calf can withstand. The results showed that the maximum peak interaction force was 161 N, which was far less than those in other studies [10, 11]. Since the robot has no joints, the reaction force of the interactive force cannot be transmitted to the ground as in study [10]. If the reaction force can be borne by the waist, the peak interaction force will be larger. A previous study has demonstrated that a large interaction force can lead to a large decrease in metabolic cost [10, 11], potentially indicating that there is still much potential for the metabolic cost to decline.

This paper used a small assisting force to achieve a large metabolic cost reduction. One advantage is that the small assisting force has little effect on gait [30]. As far as the current consensus is concerned, it is always good to not change the walking gait. After all, changing the gait will cause an increase in metabolic cost. The other advantage is that smaller motors and reducers can be used because of the improvement in efficiency. It has been reported that a 1 kg reduction in load can lead to a 1% to 2% reduction in metabolic cost [17]. Our assisting principle provides a solution to the lightweight design of active robots.

**Conclusion**

In summary, this paper proposes a new method that utilizing the energy harvest function of the Achilles tendon can improve the assisting efficiency of the active robot and designs a novel ankle robot to test the method. The initial result supports the validity of the method. This method can further reduce the active robot weight which is the
main challenge of active robots.

Moreover, some technological innovations are applied to the design of robots. The structure that separates the driver and actuator solves the problem of the Bowden cable-driven robot affecting the unassisted joints. Additionally, we propose a more accurate method that uses image processing technology to process plantar pressure data to determine the assistance timing.

The assistance method, mechanism design, hardware selection, method of assistance timing determination and control method of this paper can provide a reference for later research.

Future work
In this paper, we only performed a basic verification of the proposed assisting method, which only involves the interaction force, interaction power, metabolic cost, and kinematics, but does not involve the performance of the muscles and tendons which are the lowest and most basic layers. Future work should explore the performance of muscles and tendons before and after the assistance of a prototype to validate the proposed assisting principle.

The gait analysis system is used to judge the timing of assistance. In the future, we can develop a portable plantar pressure sensor to judge the timing of assistance so that the robot can become a real autonomous robot.

Additional file
Table S1: weight of the prototype. Table S2: mechanical power. Table S3: metabolic cost.

Abbreviations
TPU: Thermoplastic polyurethanes.

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Authors’ contributions
GH conceived study design, analyzed the data, discussed the result, and drafted the manuscript. ZW designed the ankle robot. BL and XX designed the control method. ZZ collected and analyzed the data. LX was involved in the study design. All authors read and approved the final manuscript.

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Availability of data and material
The datasets analyzed during the current study are available from the corresponding author on reasonable request.

Ethics approval and consent to participate
Informed consent was obtained from all participants to complete the protocol approved by the Guangzhou First People’s Hospital Department of Ethics Committee.

Consent for publication
The authors received consent for the publication of the photographs used within the manuscript.

Competing interests
The authors declare that they have no competing interests.

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Reference
[1] C. Tudor-Locke, W. D. Johnson, and P. T. Katzmarzyk, “Accelerometer-determined steps per day in US adults,” Med Sci Sports Exerc, vol. 41, no. 7, pp. 1384-91, 2009.
[2] Westerterp and K. R., "Physical activity and physical activity induced energy expenditure in humans: measurement, determinants, and effects," *Frontiers in Physiology*, vol. 4, 2013.

[3] A. J. Young and D. P. Ferris, "State of the Art and Future Directions for Lower Limb Robotic Exoskeletons," *IEEE Trans Neural Syst Rehabil Eng*, vol. 25, no. 2, pp. 171-182, Feb 2017.

[4] L. M. Mooney and H. M. Herr, "Biomechanical walking mechanisms underlying the metabolic reduction caused by an autonomous exoskeleton," *J Neuroeng Rehabil*, vol. 13, p. 4, Jan 28 2016.

[5] S. H. Collins, M. B. Wiggin, and G. S. Sawicki, "Reducing the energy cost of human walking using an unpowered exoskeleton," *Nature*, vol. 522, no. 7555, pp. 212-215, 2015.

[6] A. B. Zoss, H. Kazerooni, and A. Chu, "Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX)," *IEEE/ASME Transactions on Mechatronics*, vol. 11, no. 2, pp. 128-138, 2006.

[7] K. N. Gregorczyk, L. Hasselquist, J. M. Schiffman, C. K. Bensel, J. P. Obusek, and D. J. Gutekunst, "Effects of a lower-body exoskeleton device on metabolic cost and gait biomechanics during load carriage," *Ergonomics*, vol. 53, no. 10, pp. 1263-1275, 2010.

[8] H. Kawamoto and Y. Sankai, "Power assist method based on Phase Sequence and muscle force condition for HAL," *Advanced Robotics*, vol. 19, no. 7, pp. 717-734, 2005.

[9] P. Malcolm, W. Derave, S. Galle, and D. De Clercq, "A simple exoskeleton that assists plantarflexion can reduce the metabolic cost of human walking," *PloS one*, vol. 8, no. 2, p. e56137, 2013.

[10] S. Galle, P. Malcolm, S. H. Collins, and D. De Clercq, "Reducing the metabolic cost of walking with an ankle exoskeleton: interaction between actuation timing and power," *Journal of neuroengineering and rehabilitation*, vol. 14, no. 1, p. 35, 2017.

[11] B. Quinlin et al., "Assistance magnitude versus metabolic cost reductions for a tethered multiarticular soft exosuit," *Sci. Robot*, vol. 2, no. 2, pp. 1-10, 2017.

[12] R. Riener, L. Lunenburger, S. Jezernik, M. Anderschitz, G. Colombo, and V. Dietz, "Patient-cooperative strategies for robot-aided treadmill training: first experimental results," *IEEE Trans Neural Syst Rehabil Eng*, vol. 13, no. 3, pp. 380-94, Sep 2005.

[13] S. K. Banala, S. H. Kim, S. K. Agrawal, and J. P. Scholz, "Robot assisted gait training with active leg exoskeleton (ALEX)," *IEEE Trans Neural Syst Rehabil Eng*, vol. 17, no. 1, pp. 2-8, Feb 2009.

[14] A. T. Asbeck, S. M. M. D. Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *International Journal of Robotics Research*, vol. 34, no. 6, pp. 744-762, 2014.

[15] K. W. Hollander, R. Ilg, T. G. Sugar, and D. Herring, "An Efficient Robotic Tendon for Gait Assistance," *J Biomech Eng*, vol. 128, no. 5, pp. 788-791, 2006.

[16] A. D. Kuo, J. M. Donelan, and A. Ruina, "Energetic Consequences of Walking Like an Inverted Pendulum: Step-to-Step Transitions," *Exercise & Sport Sciences Reviews*, vol. 33, no. 2, pp. 88-97, 2005.

[17] R. C. Browning, J. R. MODICA, R. KRAM, and A. GOSWAMI, "The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking," *Med Sci Sports Exerc*, vol. 39, no. 3, pp. 515-525, 2007.

[18] Y. Ding et al., "Effect of timing of hip
extension assistance during loaded walking with a soft exosuit," *J Neuroeng Rehabil*, vol. 13, no. 1, p. 87, Oct 3 2016.

[19] J. Kim *et al.*, "Reducing the metabolic rate of walking and running with a versatile, portable exosuit," *Science*, vol. 365, no. 6454, pp. 668-672, 2019.

[20] G. S. Sawicki, C. L. Lewis, and D. P. Ferris, "It pays to have a spring in your step," *Exercise and sport sciences reviews*, vol. 37, no. 3, p. 130, 2009.

[21] D. J. Farris and G. S. Sawicki, "The mechanics and energetics of human walking and running: a joint level perspective," *J R Soc Interface*, vol. 9, no. 66, pp. 110-8, Jan 7 2012.

[22] M. Kaneko, T. Yamashita, and K. Tanie, "Basic considerations on transmission characteristics for tendon drive robots," in *Fifth International Conference on Advanced Robotics* Robots in Unstructured Environments, 1991, pp. 827-832: IEEE.

[23] K. J. Stuempfle, D. G. Drury, and A. L. Wilson, "Effect of load position on physiological and perceptual responses during load carriage with an internal frame backpack," *Ergonomics*, vol. 47, no. 7, pp. 784-789, 2004.

[24] C. DEVROEY, I. JONKERS, A. D. BECKER, G. LENAERTS, and A. SPAEPEN, "Evaluation of the effect of backpack load and position during standing and walking using biomechanical, physiological and subjective measures," *Ergonomics*, vol. 50, no. 5, pp. p.728-742, 2007.

[25] M. Lafiandra and E. HARMAN, "The Distribution of Forces between the Upper and Lower Back during Load Carriage," *Medicine & Science in Sports & Exercise*, vol. 36, no. 3, pp. 460-467, 2004.

[26] S. Golriz, J. J. Hebert, K. B. Foreman, and B. F. Walker, "The effect of hip belt use and load placement in a backpack on postural stability and perceived exertion: a within-subjects trial," *Ergonomics*, vol. 58, no. 1, pp. 140-147, 2015.

[27] G. A. Holloway, D. C. H., K. D., and C. J., "Effects of external pressure loading on human skin blood flow measured by 133Xe clearance," *Journal of Applied Physiology*, vol. 40, no. 4, pp. 597-600, 1976.

[28] L. M. Mooney, E. J. Rouse, and H. M. Herr, "Autonomous exoskeleton reduces metabolic cost of human walking during load carriage," *Journal of neuroengineering and rehabilitation*, vol. 11, no. 1, p. 80, 2014.

[29] F. Y. Shih, *Image Processing and Pattern Recognition*. 2010.

[30] A. T. Asbeck, S. M. M. D. Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *International Journal of Robotics Research*, vol. 34, no. 6, pp. 744-762, 2015.

[31] J. M. Brockway, "Derivation of formulae used to calculate energy expenditure in man," vol. 41, no. 6, pp. 463-471, 1987.

[32] M. P. Kadaba, H. K. Ramakrishnan, and M. E. Wootten, "Measurement of lower extremity kinematics during level walking," *Journal of Orthopaedic Research*, vol. 8, no. 3, pp. 383-392, 1990.

[33] F. A. Panizzolo *et al.*, "A biologically-inspired multi-joint soft exosuit that can reduce the energy cost of loaded walking," *Journal of neuroengineering and rehabilitation*, vol. 13, no. 1, p. 43, 2016.