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Biomechanics study of Human lower limb walking: implication for design of power-assisted robot

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Abstract—**Walking gait based on actual machine elements that capture the major features of human locomotion may enhance the understanding of human leg morphology and control, and provide necessary background science for the design and control of human-robot exoskeleton interaction system. This paper describes an investigation into the biomechanical effects of human lower limb joints during different loaded walking. Healthy volunteers walked on a level with different loaded (0kg, 10kg, 20kg, 30kg). The experiment results of human walking motion may give the base which joint of power-assisted lower extremity robot should be actuated, and what driving method should be used. At the same time, the kinematics and kinetics data of human walking motion give some bases for the choose of the actuator.**

I. Introduction

BIOMECHANICAL analysis of human walking has been increasingly utilized over the past twenty years. While most such studies have focused on gait pathology and sport technique, fewer studies applied for the design of bionic-mechanics[1]. The previous studies have shown that an improved understanding of human gait major features and human leg morphology and control in walking may shed light on more effective exoskeleton leg architectures [2-3].

The aim of our study of human lower limb joints during different loaded walking is to design an efficient, low-mass walking power-assisted lower extremity robot. Walking power-assisted robot is a kind of human-robot exoskeleton interaction system, which can be worn on to enhance human waling ability. It includes of two powered anthropomorphic legs, parallels to the human lower limbs and interfaces to the human via shoulder straps, waist belt, thigh cuffs, and a shoe connection such that the geometry of the human and the machine approximately match one another. Some power-assisted exoskeletons have been developed in the past decades [4-11], such that, Kazerooni's group in university of California, Berkeley, developed the Berkeley Lower Extremity Exoskeleton (BLEEX). In Japan, the Hybrid Assistive Leg (HAL), has been developed by Sankai et al. A quasi-passive exoskeleton concept has been advanced in the Biomechatronics Group at the Massachusetts Institute of Technology Media Laboratory, and MIT exoskeleton has been developed. Many exoskeletons among of them use active actuator, such as servo motor and hydraulic actuator, which demand a great deal of power supply to achieve system autonomy. If both passive and active elements are used in exoskeleton system, more efficient exoskeleton will achieved. MIT media Lab developed a lightweight, under-actuated exoskeleton, and the active and passive joint components of the exoskeleton were designed by examining biomechanical data from human walking [12].

In this paper, we focused on the kinematic and kinetic study of human gait during different loaded walking which based on 3D motion capture equipment combined with plantar pressure measurement device, and obtained the effects for joint angles, joint moments and powers of human lower limb with increasing load. Some interesting and potentially useful information has emerged which might lead to improvements in design of power-assisted robot.

II. Method

A. The Experiment Introduce of Human Walking Gait Measurement

We are constructing our human walking gait measurement platform. Our system composes of movement image analysis subsystem and plantar pressure measurement subsystem. Movement image system consists of four CCD cameras for data acquisition. The cameras were arranged that each target marker which placed on the key point of the subject's body remained in the field of view of every camera throughout the entire data collection, and the output of force plate was synchronized to the video data to supply the additional input required for the kinetic estimates. Then the computer analyses and calculates these original data with the biomechanics model of human, and some data which can not be measured directly, e.g. the torque and power of joint can be obtained.

It is worth specially being noted that the experiment original coordinate data of markers which placed on the key points of the subject's body and the plantar pressure used in this paper are measured through 3D motion capture system of Motion Analysis Corporation because our hardware device of experiment system is building. The experiment scene image is shown in Fig.1. According to these original data (the coordinate values of markers and three components of ground reaction forces in the anterior–posterior, medial–lateral, and vertical directions while three moments) and the human lower

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limb model, the kinematics and kinetic of walking gait were outputted by our software system which we have developed with C# and Matlab language.

Fig.1 Experiment scene image without load walking

B. Human Lower Limb Modeling

In this study, the human lower limb includes the thigh, calf and foot has been adopted, as shown in Fig.2. All body segments are assumed to be rigid and their motions are modeled in the sagittal plane only. An inverse dynamics approach has been adopted [13], where the measured motions of all the body segments of lower limb, are given as the input data, and the ground reactions are measured using force plates and inputs to the calculation. Then basing the inverse dynamics on measured kinematics data, integrating body segment parameters, centers of gravity, angular kinematics and ground reaction forces, and applying Newton's second law of motion to each segment, the joint forces, moments and powers in the sagittal plane can be obtained.

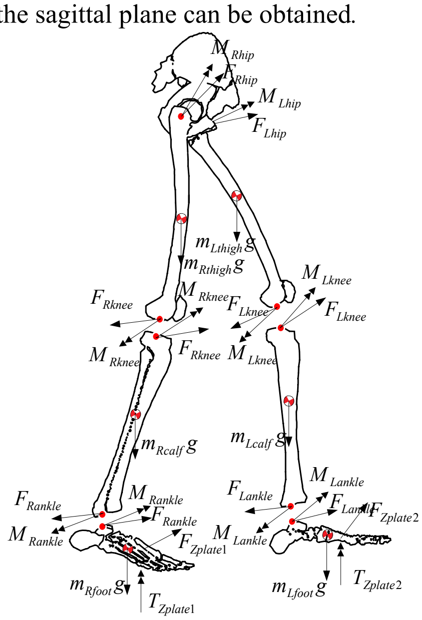


Fig.2 The human lower limb model

The equations of right foot motion in the sagittal plane, can be expressed as

$$F_{Rankle} + m_{Rfoot} g + F_{plate1} = m_{Rfoot} a_{Rfoot}$$

$$\dot{L}_{Rankle} = M_{Rankle} + T_{Zplate1} + M_{FRankle} + M_{FR}$$

Where, a_{Rfoot} is the acceleration of right foot CG, $\dot{L}_{Rankle}$ is rate of change of angular momentum of the right ankle segment, $M_{FRankle}$ is moment due to F_{Rankle}, M_{FR} is moment due to F_{plate1} .So the joint force and moment of right ankle which F_{Rankle}, M_{Rankle} can be calculated with equations (1) and (2)

$$F_{Rankle} = m_{Rfoot} a_{Rfoot} - m_{Rfoot} g - F_{plate1} \tag{1}$$

$$M_{Rankle} = \dot{L}_{Rankle} - T_{Zplate1} - M_{FRankle} - M_{FR} \tag{2}$$

Then the joint power of right ankle P_{Rankle} can be calculated with equation (3)

$$P_{Rankle} = M_{Rankle} \bullet \dot{\theta}_{Rankle} \tag{3}$$

Where, $\dot{\theta}_{Rankle}$ is the joint angle velocity of right ankle.

The same for the right calf, so the joint force, moment and power of right knee can be calculated with equations (4), (5) and (6).

$$F_{Rknee} = m_{Rcalf} a_{Rcalf} - m_{Rcalf} g - F_{Rankle} \tag{4}$$

$$M_{Rknee} = M_{FRknee} - M_{FRankle} - \dot{L}_{Rcalf} \tag{5}$$

$$P_{Rknee} = M_{Rknee} \bullet \dot{\theta}_{Rknee} \tag{6}$$

Where, a_{Rcalf} is the acceleration of right calf CG, $\dot{L}_{Rcalf}$ is rate of change of angular momentum of the right calf segment, $M_{FRankle}$ is moment due to F_{Rankle}, M_{FRknee} is moment due to F_{Rknee}, $\dot{\theta}_{Rknee}$ is the joint angle velocity of right knee.

The joint force, moment and power of right hip can be obtained with equations (7), (8) and (9).

$$F_{Rhip} = m_{Rthigh} a_{Rthigh} - m_{Rthigh} g - F_{Rknee} \tag{7}$$

$$M_{Rhip} = M_{FRhip} - M_{FRknee} - \dot{L}_{Rthigh} \tag{8}$$

$$P_{Rhip} = M_{Rhip} \bullet \dot{\theta}_{Rhip} \tag{9}$$

Where, a_{Rthigh} is the acceleration of right thigh CG, $\dot{L}_{Rthigh}$ is rate of change of angular momentum of the right thigh segment, M_{FRknee} is moment due to F_{Rknee}, M_{FRhip} is moment due to F_{Rhip}, $\dot{\theta}_{Rhip}$ is the joint angle velocity of right hip.

III. THE EXPERIMENT RESULTS

Five students (males) without any known neurological or musculoskeletal disorders participated in the experiments. The mean age of subjects was 25±2 years, mean body mass was

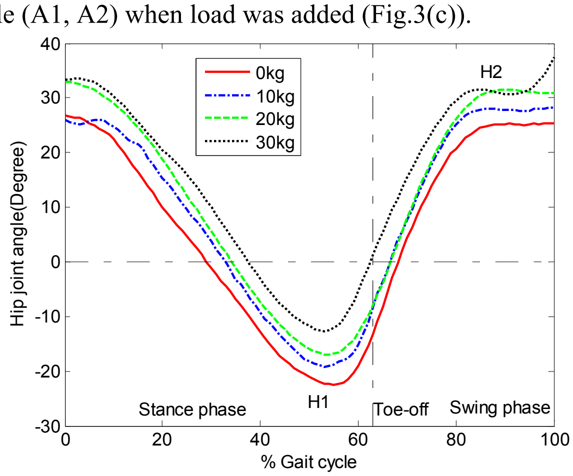
57±6.5kg, and mean height was 1.7±0.03m. Each subject was fitted with 22 retroreflective markers placed over key points of the body, and subject walked with different load (0kg, 10kg, 20kg and 30kg) walking on force platform concealed in a flat walkway, while motion data of markers was collected using cameras and three components of ground reaction forces (in the anterior–posterior (F_x), medial–lateral (F_y), and vertical (F_z) directions) while three moments (M_x, M_y, and M_z) were measured synchronously. The ground reaction force data were sampled at 1200 Hz and kinematics data were captured at 60 Hz.

All data were synchronized. All kinematics and kinetic data were time-normalized to 100 points over the stride (62 for stance, 38 for swing). The values for each point of interest was taken from all trials and averaged within subjects for each experiment condition.

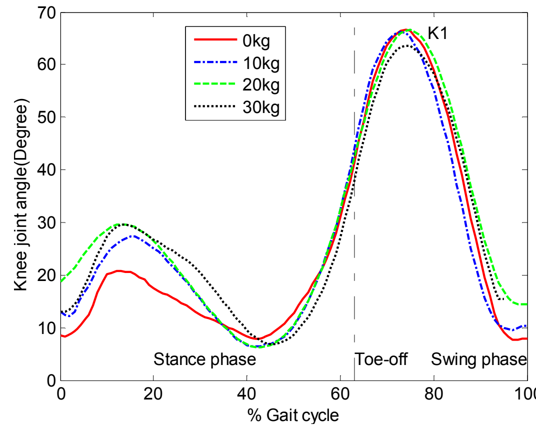
A. Joint kinematics

The sagittal plane is the dominant plane of motion during human locomotion. So, this paper only gives the human joint motion data during different loaded walking in the sagittal plane.

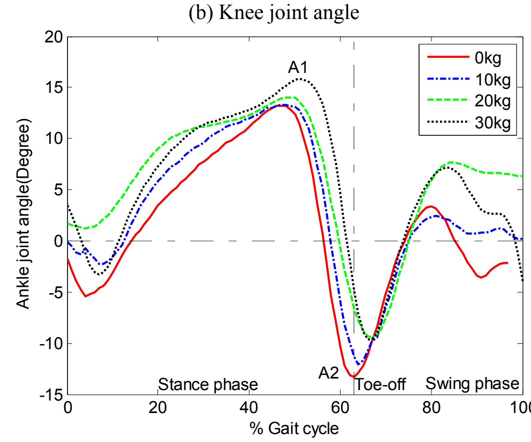
Joint kinematics data in different loaded walking are presented in Fig.3. The hip, knee and ankle joint angle curves in different loaded walking are not significantly difference, that is, the human kinematics are not substantially altered when walking with load increasing. Fig.3(a) indicated that the maximum hip extension (H1) angle decreased as the load increase, but the maximum hip flexion angle (H2) increased as the load increase. Fig.3(b) showed that 0kg and 10kg load effect for the maximum knee flexion angle (K1) was not significant may be due that 10kg load is light for subjects, and the maximum knee flexion angle did not start to decrease until the 30kg load was carried. The ankle showed a trend (although not significant) for an increase in the maximum dorsiflexion angle (A1, A2) when load was added (Fig.3(c)).



(a) Hip joint angle



(b) Knee joint angle

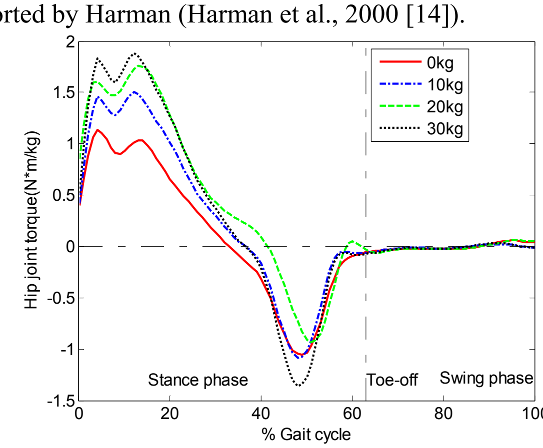


(c) Ankle joint angle

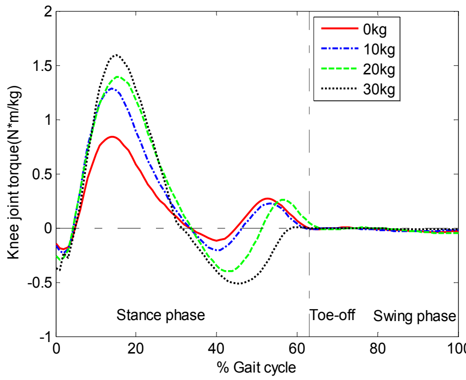
Fig.3 Joint angle in different loaded walking

B. Joint Moments and Powers

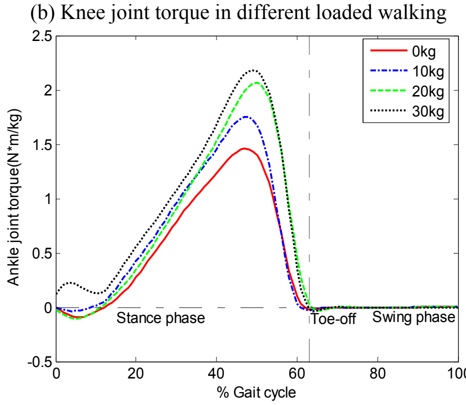
Fig.4 and Fig.5 give the joint moments and powers in different loaded walking. The results showed torques and powers about the ankle, knee and hip increased with load. we can found that when the body plus mass increase by 17.54%, hip peak extension torque increased about by 19.37%, knee peak extension torque increase about by 41.28%, ankle peak plantar flexion torque increased about by 15.1%, and these results almost are consistent with experimental studies reported by Harman (Harman et al., 2000 [14]).



(a)Hip joint torque in different loaded walking

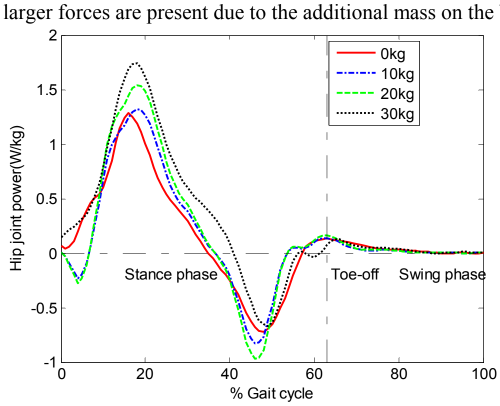


(b) Knee joint torque in different loaded walking

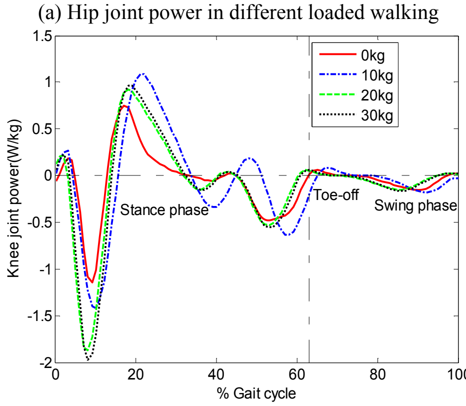


(c) Ankle joint torque in different loaded walking

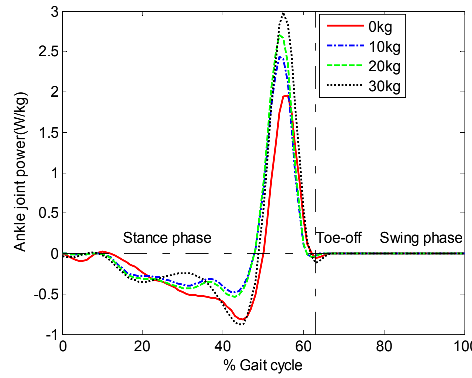
Fig.4 Joint torques in different loaded walking, and torques are normalized by body mass. Joint torques are larger with loaded increasing, which is expected since the larger forces are present due to the additional mass on the back.



(a) Hip joint power in different loaded walking



(b) Knee joint power in different loaded walking



(c) Ankle joint power in different loaded walking

Fig.5 Joint powers in different loaded walking, and powers are normalized by body mass. Joint powers are larger with loaded increasing, which is expected since the larger forces are present due to the additional mass on the back

IV. THE HUMAN WALKING MOTION ANALYSIS AND IMPLICATIONS FOR DESIGN OF POWER-ASSISTED ROBOT

The study of human walking gait analysis system can provide the joint torques and power data required for walking, these can give some bases for the choose of actuator. Once the actuators sizes and the mounting positions were chosen to ensure the required range of motion and required torque, the joint velocity data was used to determine the actuator rotate speed required to walk, and the human power required for walking is important for selecting the power source of robot.

A conservative estimate of the weight of the exoskeleton and load in backpack was to be about 65kg and the kinematics and kinetics data were scaled to a 65kg person in order to estimate the torques and powers required at the joints of the power-assisted robot.

A. The Hip Joint

From Fig.3(a) it can be seen that during level walking, the human hip joint follows an approximate sinusoidal pattern, and the range of motion of the hip joint vary between -25 to 38 degrees. The hip torque in Fig.4(a) has both positive and negative, hence a bi-directional hip actuator is required. Hip torque is positive in stance phase as the hip supports the load on the stance leg and propels the leg forward. In late swing, the torque gets smaller as the hip decelerates the leg prior to heel strike. Hip power in Fig.5(a) has both positive and negative during walking cycle, but the average hip power is positive, indicating an actuator is required at the hip, so in the design of power-assisted robot, an active actuator is need in the flexion/extension motion. However, the motion range in hip abduction/adduction and internal/external rotation direction is small, so hip abduction/adduction is installed with a linear spring, using a spring capturing the negative power and release it during the positive power period to realize the motion requirement and satisfy the walking comfort, and hip rotation joint is completely free.

According to the experiment results for the design of hip joint, a summary of specifications for human walking gait data are shown in Table I. These data can give some bases for the choose of power-assisted robot hip joint actuator.

Table I Hip joint specifications were extracted from the human walking gait data (The values was taken from averaged within subjects (body mass was about 65kg, walking speed was about 1.3m/s))

Parameter	Value
Range of motion	-25^0-38^0
Max joint velocity	2.25rad/s
Max joint torque	92Nm
Max joint power	95W

B. The Knee Joint

Fig.3 (b) shows that an average person can flex their knee from 8^0 to 68^0 during level walking. The knee torque (as shown in Fig.4 (b)) has both positive and negative, indicating the need for a bi-directional actuator. Fig.5 (b) gives the knee power result in different loaded walking. The average power of knee joint is negative, mostly dissipates energy throughout the walking cycle and is best suited with a dissipative device (i.e., damper) located at this joint. It can be also seen from Fig.5 (b), that during early stance, the knee behaves like a spring as there is a region of negative energy followed by a region of positive energy of similar size. For the remainder of the gait cycle, the knee acts like a variable-damper to control leg during the swing phase. However, the study shows that [15], the knee requires a large amount of positive power when the human walking up steps, an incline, or squatting. So the knee joint is actuated or not should be confirmed according to the function require of device. If the power-assisted robot is mainly applying for walking strength enhancement over a long distance or load augmentation to carry heavy load, the knee joint requires a driver, but if the power-assisted robot is only applying for disorder persons or aged people walking in flat floor, a spring and damper are best suited.

Table Ⅱ gives knee joint specifications of human walking gait data, and according to the experiment results for the knee joint design of power-assisted robot.

TableⅡ Knee joint specifications were extracted from the human walking gait data(The values was taken from averaged within subjects(body mass was about 65kg, walking speed was about 1.3m/s))

Parameter	Value
Range of motion	-8^0-68^0
Max joint velocity	3.29rad/s
Max joint torque	58Nm
Max joint power	78W

C. The Ankle Joint

Fig.3(c) shows a small ankle range of motion during level walking, approximately -14^0 to 15^0. Fig.4(c) shows the ankle torque is almost entirely negative, indicating unidirectional ankle actuator is required. However, if symmetric bi-directional actuator is considered, spring-loading would allow the use of low torque producing actuators [16].The ankle mechanical power in Fig.5(c) has also both positive and negative during walking cycle. Previous study showed that if a period of negative power precedes a period of positive power than it may be possible to capture the negative power in a spring and release it during the positive power period, so the spring may be implemented at these joints for reduced power [12]. Fig.5(c) also shows that the region of negative is approximately equal to the region of positive power during without load walking, but with the load increasing, the region of positive power is significantly larger indicating that an actuator is required for ankle joint. However, a heavy actuator at the ankle would be detrimental to walking metabolism due to the large distal mass.

We also find that the region of positive power is larger than negative with walking speed increasing. Figure 6 gives the plot of ankle angle versus ankle torque for walking at three speeds (slow walking 0.8m/s, normal walking 1.3m/s and fast walking 1.7m/s). Ankle moment versus ankle angle curves displayed hysteresis loops at different walking speed [17,18]. The area inside the hysteresis loops was reduced with the speed decrease, and was near aero for slow walking. A linear relationship can be seen during the slow walking and the torsional spring constant is about 357Nm/rad. A hybrid actuation approach where in ankle joint a motor is used in conjunction with the spring may be suited in case of at faster walking speed or with larger load walking. Table Ⅲ gives ankle joint specifications of human walking gait data. These data are important for the ankle joint actuator of power-assisted robot.

Table Ⅲ Ankle joint specifications were extracted from the human walking gait data (The values was taken from averaged within subjects(body mass was about 65kg, walking speed was about 1.3m/s))

Parameter	Value
Range of motion	-14^0-15^0
Max joint velocity	2.47rad/s
Max joint torque	97.5Nm
Max joint power	116W
Torsional Spring constant	357Nm/rad

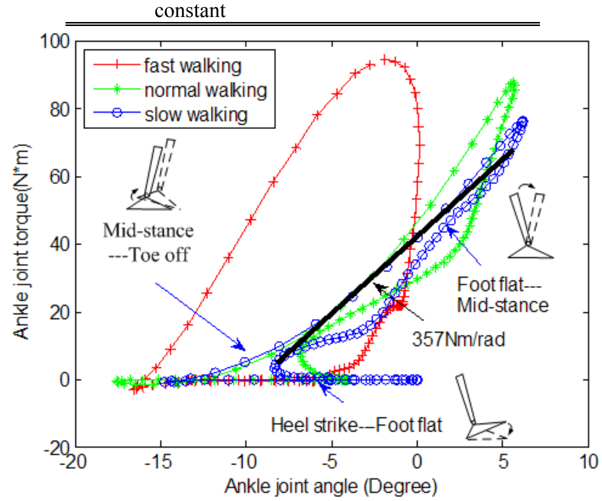


Fig.6 Plot of ankle angle versus ankle torque for the walking cycle at three walking speeds (fast walking 1.7m/s, normal walking 1.3m/s and slow walking 0.8m/s). The segment 1-2 represents the ankle torque-angle behaviour during plantarflexion which begins at heel-strike and ends at foot-flat, the segment 2-3 represents the ankle torque-angle behaviour during dorsiflexion which begins at foot-flat and continues until the ankle reaches a state of maximum dorsiflexion, and 3-4 represents the ankle torque-angle behaviour during powered plantarflexion which begins at mid-stance and ends at the instant of toe-off.

V. CONCLUSIONS

The biomechanics of human walking gait were studied in this paper, and this study has answered the question of which joint of powered-assisted robot should be actuated and what driving method should be used. In collusions as following:

1) The human hip joint torque in flexion/extension direction during level walking has both positive and negative, hence a bi-directional hip actuator is required. At the same time, hip power has both positive and negative during walking cycle, but the average hip power is positive, indicating an actuator is required at the hip. The implication of human hip motion in the design of power-assisted robot is such that, an active actuator is required for the hip joint flexion/extension motion.
2) The human knee joint torque in flexion/extension direction has both positive and negative, indicating the need for a bi-directional actuator. The average of human knee joint power during level walking is negative, mostly dissipates energy throughout the walking cycle and is best suited with a dissipative device (i.e., damper) located at this joint. But when the power-assisted robot is mainly applying for carry heavy load during different walking conditions, such as walking up steps, an incline, or squatting, an active actuator is necessary for power-assisted robot knee joint.
3) Human ankle joint torque in flexion/extension direction is almost entirely negative during level walking, indicating unidirectional ankle actuator is required. However, if symmetric bi-directional actuator is considered, spring-loading would allow the use of low torque producing actuators. The ankle mechanical power has also both positive and negative, and negative power precedes a period of positive power during walking cycle, so the spring may be implemented for reduced power. Our experiments also showed that the region of positive power is larger than negative with walking speed or load increasing, indicating that an active actuator is required for ankle joint. However, a heavy actuator at the ankle would be detrimental to power consuming and walking comfortlessness due to the large distal mass. So a hybrid actuation approach which combined active and passive elements, such as a motor is used in conjunction with the spring may be suited for ankle joint in case of at faster walking speed or with larger load walking.
4) The human walking biomechanics study can provided the joint torques and power data required for walking, these can give some bases for the choose of actuator. Once the actuators sizes and the mounting positions were chosen to ensure the required range of motion and required torque, the joint velocity data was used to determine the actuator rotate speed required for walking, and the human power required for walking is also important for selecting the power source of robot.

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