Designing of a Passive Knee-Assisting Exoskeleton for Weight-Bearing

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Abstract. Weight-bearing exoskeleton can effectively help the wearer to bear heavier burden, while assisting his ambulation. However, current researches in this field are relatively scarce on the passive weight-bearing exoskeleton. This research aims to design an unpowered knee-assisting exoskeleton used for weight-bearing, which can store human metabolic energy and assist human locomotion, through utilization of Teflon string, pulley, and compression spring. Biomechanical analysis of human weight-bearing locomotion shows that knee-flexion angle can be used to identify level walking or ascending movement of a person, where the largest sagittal flexion angle in level walking does not exceed 60º. Hence, the contour of pulley used for winding string is designed to be eccentric, in which the assisting torque varies nonlinearly according to the knee-flexion angle. Through mechanical modeling of eccentric pulley, we predict the assisting torque of the device and compare it with actual experimental statistics. Results show that such passive exoskeleton exhibits multi-stage nonlinear assisting augmentation under different knee-flexion angles; with least possible knee assistance during level walking, and remarkable assistance during climbing.

Keywords: Weight-bearing · Knee-assisting · Unpowered · Ascending locomotion · Eccentric pulley

1 Preface

Exoskeleton is a wearable mobile device that can amplify the strength or save the energy consumption of an individual in motion, through the use of external power supply or human power [1, 2]. In recent years, exoskeleton technology has been used in various fields such as load-carrying, rehabilitation of paraplegics [3], human locomotion, object-lifting, and industrial manufacturing. The main goal of load-carrying

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exoskeleton is to enhance strength, reduce fatigue, as well as protect musculature damage of a healthy wearer [4]. The basic functions of such device are for weight support, gait assistance, while ensuring the wearer's flexibility in undesirable road conditions, and retaining long-lasting working time.

The Berkley's BLEEX [5, 6], Harvard's Soft Exosuit [7, 8], Japan's Hal [9], and France's Hercule [10] are some typical active powered exoskeletons made to decrease fatigue and increase productivity. Though these machines greatly improved the wearer's capabilities of the natural human skeletal structure, there are some limitations like bulky size, high power consumption (e.g. the power sources can only sustain one exoskeleton for no longer than a few hours), complicated systems, expensive research and production costs, have constrained the machine's applications and promotions. In contrast to powered exoskeleton, passive exoskeleton has simpler structures, lower production costs, and requires no electric power at all, thus making it more practical and promotable than active exoskeletons.

Lockheed Martin's Fortis is a representative example of passive exoskeleton [11]. It employs linkage mechanism and dead-center principle, which gives the Fortis a higher productivity than ever before. This unpowered framework allows the wearer to effortlessly lift heavy tools. Unfortunately, due the design's lack of strength-amplification feature, Fortis is more useful in standing position, and not for outdoors weight-bearing activities.

In fact, not only can powered exoskeleton provide weight support for the wearer, but also can passive exoskeleton. For instance, Cameron University designs an ankle exoskeleton using techniques like strings, stretch springs, and ratchet [12, 13]. The passive ankle elastic mechanism can be used to produce ankle exoskeleton that enables the wearer to walk with minimal actuation. Such kind of purely passive exoskeleton can reduce the walker's metabolic energy by up to 7%.

Over centuries of evolution, human beings have already adapted well-coordinated mobility. During walking, dead-centre principle explains why the knee stands upright; pendulum effect is observed when the walker's legs swing freely without bearing weight of the torso on sagittal plane. This manner of walking permits humans to walk on flat roads at higher efficiency with lower energy consumption [14]. Therefore, the unpowered exoskeleton developed by Cameron University marks the beginning of humans' attempt to optimize human locomotion through exoskeleton.

Cameron University focuses on the development of ankle exoskeleton used for gait assistance, which is somewhat different from weight-bearing exoskeleton. As a person ascends (climbing uphill or walking upstairs), his quadriceps are constantly doing work, thus, offering knee torque for its extension. Such torque overcomes gravitational potential energy exerted on the walker's own body weight and his loads. If the human body ascends while bearing excessive weight, it would not take long before he drains out his energy in the lower limbs, resulting in exercise-induced fatigue and even exercise-induced exhaustion.

Hence, this exoskeleton not only need to provide effective weight-bearing support for the wearer both in standing position and in motion, but also to provide additional knee torque when ascending, in order to reduce output power of prime mover muscles (e.g. quadriceps), and ultimately to prevent weariness from overworking the leg muscles. As a result, our team will concentrate on passive knee joint exoskeleton development. We will use springs to collect the wearer's metabolic energy, provide added knee-assisting torque while he walks upwards, hoping to put this purely passive weight-bearing exoskeleton technology to use.

2 Human Weight-Bearing Gait Biomechanics Analysis

In order for a purely passive exoskeleton to provide supplementary strength for the wearer, it must obtain energy from the wearer first. To simplify the structure's complexity, the best way for the knee-assisting device to store energy should also come from the knee joint movement. When the human body is walking (on flat road, uphill, or downhill), the knee joint movement involves both extension and flexion of the legs. For every one step upward, the leg stepping forward first will experience leg flexion and then leg extension when it brings the body upwards. Thigh experiences fatigue mainly because when the leg extends to prepare for rising, the quadriceps need to overcome gravitational potential energy to do work. If an assisting device is present, it can store the energy that would otherwise be used for leg flexion when moving upward. This device works in that the hamstring tendon uses more than needed energy to pull the strings of the device when the wearer bends leg for rising, the stored energy is later released when the wearer rises. The goal of this process is to distribute the wearer's metabolic energy between leg flexion and extension, in order to prolong work period throughout the step going upward. Even though this approach cannot reduce the total amount of work required, it could reduce the maximum burden on the wearer, hence increases the total time of weight-bearing ambulation that the wearer can endure.

Meanwhile, the human body also experiences leg extension and flexion in level walking to avoid the legs from hitting the ground when moving forward. Here, the knee-assisting device should reduce the effect of weight assistance and energy storage, or it would interfere with normal human ambulation and cause unnecessary energy consumption of the wearer. Therefore, our device needs to function differently between level walking and climbing up; to provide visible assistance for the wearer when going upward, and minimize intervention in the wearer's natural marching on level ground. As a purely passive device with no electrical monitors, the best alternative to distinguish level walking and climbing up movements is by looking at the knee joint flexion angle, where the angle is considerably bigger in level walking than in climbing up.

To find the most suitable knee joint flexion angle to distinguish level walking and ascending locomotion, we conducted a "maximum knee-flexion angle" statistical test for the human body under different conditions. During the experiment, we placed the camera axis perpendicular to wearer's sagittal plane, and videotaped the wearer's movements where the lower limbs stay in the center of the video. We marked the measuring points on the thigh and calf to precisely measure the respective flexion angles. Figure 1 below displays pictures from a previous experiment.

During this experiment, the measurable value is the maximum knee-flexion angle, and factors that could affect experimental results include wearer's gait and whether he carries any load while in motion. (See Table 1 for details.) To derive data of significant statistical meaning, we conducted the same experiment on 5 individual samples, in which every sample was tested for 10 times under each different experimental

Fig. 1. Maximum knee-flexion angle statistical test for level walking

environment. Eventually, we obtained the statistical mean and standard error of maximum knee-flexion angle for every individual sample and under different experiment conditions. The experiment results shown in Table 1 are the average level of maximum knee-flexion angle of all samples, with the standard error among different individuals following behind.

Table 1. Maximum knee-flexion angle statistical test for various locomotion

Locomotion		0 kg	10 kg	30 kg
Level walking	Stroll (1.36 m/s)	59.5 ± 1.9°	59.9 ± 1.8°	64.8 ± 1.3°
	Speed walk (1.59 m/s)	62.9 ± 2.1°	63.5 ± 2.0°	67.9 ± 1.5°
	Jog (2.55 m/s)	77.4 ± 2.1°	78.3 ± 1.8°	81.4 ± 2.4°
Step ascending	Bump (10 cm)	67.0 ± 1.9°	68.6 ± 1.7°	72.8 ± 1.4°
	Stair (20 cm)	80.9 ± 1.4°	81.1 ± 1.6°	89.6 ± 2.2°
	Step (30 cm)	98.8 ± 1.7°	101.6 ± 1.7°	112.0 ± 2.1°
	Bleacher (40 cm)	109.6 ± 2.0°	112.0 ± 1.7°	118.3 ± 2.1°

From Table 1 above, we can see that, on average, human level walking flexion angle is 59.5 ± 1.9°. This range is correspondent to the CGA's flexion angle of 58.2°. In contrast, human ascending flexion angle is about 67.0–118.3°. For example, Fig. 2 below shows a sample's maximum flexion angles when climbing up different altitudes.

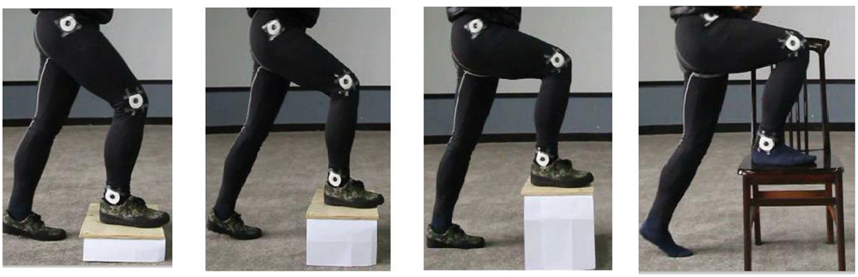

Fig. 2. Maximum knee-flexion angle at different heights

As the inclination altitude increases, the human knee-flexion angle can become greater than 118.3°. Under normal conditions, such manner of walking would no longer be single average strides, but of steep-slope climbing. Therefore, knee joint weight-assisting device should be designed in which it provides most effective weight-assistance at flexion angle of 60–120°, and lowest possible weight-assistance at flexion angle of 0–60°.

3 Passive Knee-Assisting Device Design

3.1 Primary Design Objectives

a. Elastic potential energy storage during knee flexion; torque assistance during extension;
b. Insignificant torque assistance during level walking, and apparent torque assistance during ascending, determined by flexion angle.

3.2 Structural Design

One applicable design is to use pulley, Teflon strings, and compression springs to actualize our exoskeleton's passive knee-assisting function. Our exoskeleton is worn along outer human leg, with one rod placed on the outer thigh and another on the outer calf, and knee-joint components located on the same height as the wearer's natural knee-joint. (As shown in the dotted lines in Fig. 3a). Figure 3b displays a profile of our exoskeleton, essential components include the shank exo-caput (1) (the exoskeleton component which connects the shank rod to the knee joint), thigh exo-caput (2), compression spring (3), Teflon string (4) with lead tails (401, 402) on both ends, slider block (5), rotation shaft (6), and thigh rod (7). Among these, both the shank exo-caput and thigh exo-caput are equipped with a channel (101, 201) where the Teflon string penetrates, and ends of the channel are the string outlets (102, 202) in which the string gets out from exo-caput's channel. Specifically, the string outlet on the thigh exo-caput is named as upper outlet (202), and the corresponding string outlet on the shank side is named as lower outlet (202). The partial pulley of the shank exo-caput is also equipped with string grooves (103). In the exo-thigh sector, the compression spring (3) sits between the slider block (5) and thigh exo-caput (2). As Teflon string passes the inside of compression spring, it then penetrates through the string channels and outlets of both exo-caputs, and eventually fastens the lead tail onto the end surface of the shank exo-caput.

During leg flexion, the shank and thigh exo-caput revolves around the rotation shaft respectively. Because total length of the Teflon string remains constant, the length of string between two exo-caputs continue to expands during leg flexion, and gradually coils up inside the string grooves. Consequently, the string segment located inside the thigh rod shrinks, causing it to tag along the slider block and press against the compression spring, and ultimately enables the device to store energy when the leg flexes. In fact, the shank exo-caput is an incomplete pulley, and one side of the string tails fastens on this incomplete pulley, thus forming a winding device. Such design allows the energy-storing spring to be built in the thigh rod of the exoskeleton, which in return

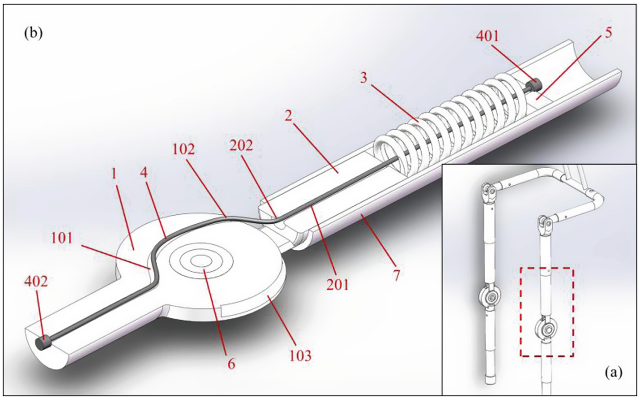

Fig. 3. Passive knee-assisting structure illustration

reduces knee-joint complexity and, increases space utilization. What is more meaningful is that the relative motion between the string and the pulley's surface is purely rolling. They won't slide against each other during locomotion, which makes the device more durable.

In leg extension, the compression spring brings the Teflon string back into the thigh rod, and induces assisting torque to shank exo-caput. Accordingly, this assisting torque (T_a) is determined by the spring force (F_s) and the force lever (h_l), where force lever is the distance between the center of rotation shaft and tangent line of the eccentric curve, shown as blue dashed line in Fig. 4a. Essentially, for the unpowered knee-assisting device to have least productivity in level walking and maximum productivity in ascending locomotion, the assisting torque (T_a) must be a nonlinear function of the knee flexion angle θ. More concretely, the function increases monotonically with the increasing of θ. However, when $\theta < 60^{\circ}$, the incremental slope is relatively small, and the slope increases when $\theta < 60^{\circ}$. A nonlinear assisting torque function can be achieved by using a nonlinear spring, varying the deformation of spring (Δs) or the arm of string force (h_l) nonlinearly from change in θ. Indeed, we can use nonlinear spring to obtain desired assisting behavior, but doing so would increase the level of difficulty in design and manufacturing. In contrast, to change the force lever (h_l) (Fig. 4a, blue dashed line segment) and string extension (s_d) (Fig. 4a, green dotted line segment), would only require making the pulley contour radius (R_k) to be an eccentric pulley that changes with the knee flexion angle θ. Such newly designed apparatus is simple and reliable, and easier to modify the mechanical characteristic for different knee-assisting devices. For this reason, our research will utilize eccentric pulley to actualize nonlinear knee-assistance.

It must be noted that, when pulley contour curve becomes eccentric, the center of curvature for any points of the pulley contour do not necessarily match with the rotation center of the knee-joint. Thus, the distance (R_k) between one edge point and the rotation center, does not necessarily equal to the distance (h_l) between the tangent line crossing such edge point and the rotation center. One example of such situation can be seen in Fig. 4b. The geometric shape of the eccentric pulley can be determined by

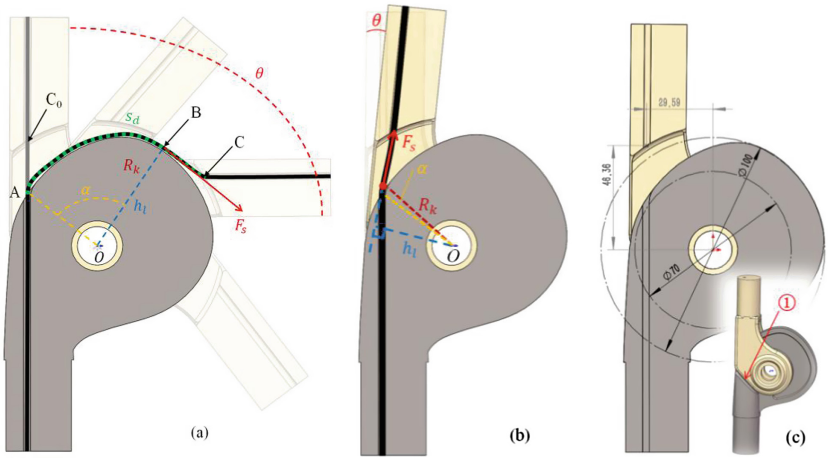

Fig. 4. Eccentric pulley parameter definition (Color figure online)

the contour's included angle (α) and contour radius (R_k). However, in order for our derivations to have actual meanings, we should set knee-joint leg flexion angle θ as the independent variable when deriving the string extension length (s_d) or string force lever (h_l). The offset distance between the upper outlet (Fig. 3b, 202) and rotation shaft, as well as the eccentric contour figure, all contribute to the nonlinear relation between flexion angle θ and the contour's included angle α, resulting in $\theta \neq \alpha$.

The outline of the eccentric pulley employed in this research is based on two concentric circles with different diameters (70 mm and 100 mm, respectively) and smoothed crossover with spline curves, as shown in Fig. 4c. Concurrently, for the knee-joint exoskeleton to retain angle-restricting function, we offset axis of the shank and thigh rods against the knee-joint rotation center, as well as set up restricting blocks on both sides of the eccentric pulley. When the exoskeleton is in upright position, shank and thigh exo-caputs' restricting blocks touch each other (Pointed as ① in Fig. 4c), thus avoiding knee-joint to bend forward, which could damage the human knee joints. Such design will also cause the upper outlet offsetting certain distance in regard to the rotation center of knee-joint (horizontal offset 29.59 mm, vertical offset 46.38 mm).

3.3 Mechanical Modeling

In order to analytically calculate the torque of such knee-assisting device, we need to obtain the analytical formula of the eccentric spline contour curve first. By using Matlab's image recognition, the numerical solution of eccentric curve (shown in Fig. 5a) can be extracted from the cutaway view of exoskeleton's knee-joint, which are plotted in Solidworks. Thus, the analytical formula of the eccentric contour curve can be derived and replotted in a polar coordinate, which can be found as a blue thick line in Fig. 5b, by using the polynomial fitting to the previous numerical solution. Those

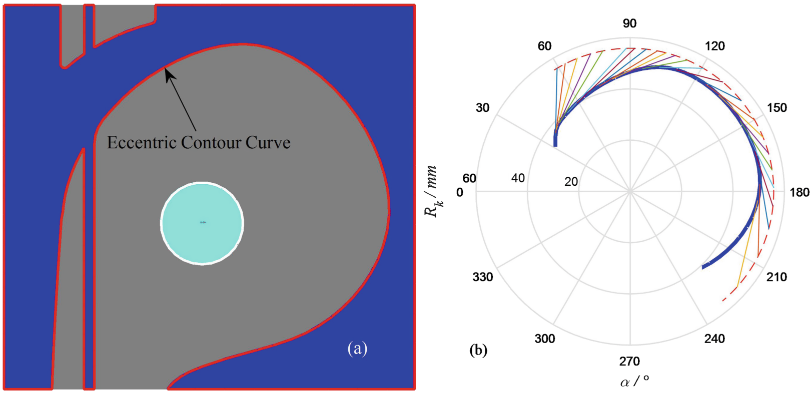

Fig. 5. Image recognition and tangent calculation of the eccentric contour curve (Color figure online)

colorful straight segments around the contour curve are the tangent lines with different contact points on that curve, which are calculated by linear fitting the slope of its differential part. Each segment's length represents the impending Teflon string between two exo-caputs, and the other ending of the segment implies the outlets of the thigh exo-caput.

According to the offset position of the upper outlet as compared to the rotation center, the rotating trajectory of the outlet can be predicted with the motion of knee flexion, which is shown as a red dotted arc in Fig. 5b. By merging formulae of such trajectory with those of different tangent lines, the variation of the string's stretching length (s_d) and the position of upper outlet (shown as C point in Fig. 4a) can be derived against the contour's included angle (α). The angle between the upper outlet's present (C point) and original position (C_0 point in Fig. 4a) indicates the flexion angle of knee joint (θ), which further gives the nonlinear relationship between θ and α shown in Fig. 6. It should to be noted that the original contour's included angle (α_0) is nonzero, which can be confirmed in Fig. 5b, hence Fig. 6 annotates a reference line ($\alpha - \alpha_0$) for a better observation of nonlinearity.

The arm of the string's force (h_l) can be calculated by the tangent line of the eccentric contour at the string's contact point (for instance, point B in Fig. 4a). In Cartesian coordinates, the distance from the rotation center (x_o, y_o) to the tangent line $Ax + By + C = 0$ is obtained as the following formula:

$$h_l = \frac{Ax_o + By_o + C}{\sqrt{A^2 + B^2}} \tag{1}$$

Hence, we retain the analytical result about the force lever of string (h_l) corresponding to the knee-flexion angle, which is shown in Fig. 7a. Obviously, the arm of force keeps increasing nonlinearly when $\theta < 60^\circ$, while maintaining the level around 50 mm when $\theta < 60^\circ$.

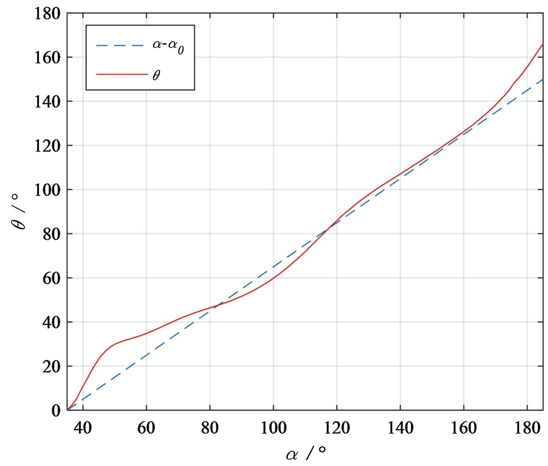

Fig. 6. The nonlinear relationship between knee flexion angle θ and contour's included angle α

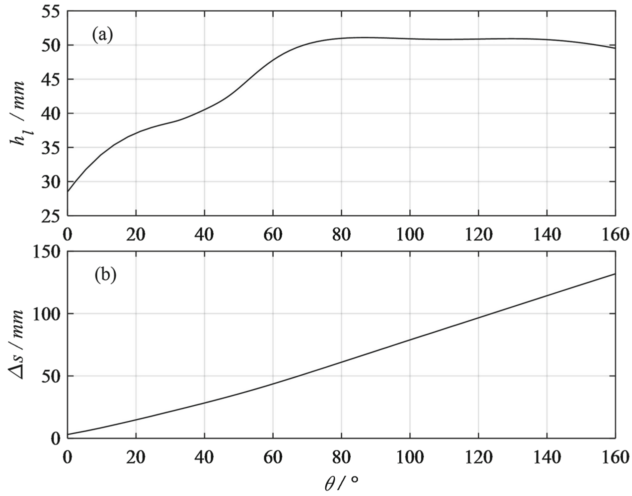

Fig. 7. Variations of string's force lever (a) and the deformation of spring (b)

In addition, the deformation of spring Δs can be obtained by the stretch length of string (s_d) between two outlets when the knee joint flexes by θ (shown as $\widehat{AC}$ Fig. 4a), subtracting its original stretch length (s_{d0}) when the knee keeps upright, $\Delta s = s_d - s_{d0}$. The length of curve $\widehat{AC}$ is composed of the arc length ($s_{\widehat{AB}} = \int_0^{\alpha} R_k(\varphi)d\varphi$) where Teflon string coils around the groove on shank exo-caput, and the distance of the impended straight string ($\overline{BC}$). Whereas the original stretch length of string (s_{d0}) can be obtained by calculating the distance of $\overline{AC_0}$. Result shows the deformation of spring Δs caused by winding the string around such eccentric pulley has a highly linear increase with the flexion angle of knee θ, which is illustrated in Fig. 7b. Furthermore, when we use a linear spring whose stiffness is k and preload sets as s_0, it will give the string a stretch force $F_s = k(\Delta s(\theta) + s_0)$ with the bending angle of knee-joint equals θ. Finally, the extension torque T_a provided by the passive knee-assisting device can be derived by the folllowing formula:

$$T_a(\theta) = F_s \cdot h_l = k \cdot (\Delta s(\theta) + s_0) \cdot h_l(\theta) \tag{2}$$

4 Distinctive Mechanical Assistance Characteristic Verification Experiment

We use Nylon 3D printing technology to fabricate the exoskeleton's knee-joint designed in the previous section; aluminum-alloy tube for the hip, shank rod, and thigh rod components; and carbon fiber sheet for backboard. Because the spring mentioned above needs to be placed inside the thigh rod, its outer diameter need to be less than the rod's inner diameter (24 mm), and the preloaded spring length needs to be less than the rod length (240 mm). Meanwhile, the exoskeleton needs to provide appropriate torque during climbing, and does not affect the wearer's normal ambulation. Through repeated trials, we chose 2.5 mm wire diameter, 23 mm outer diameter, 230 mm length, 20 coils of compression spring with approximate stiffness of 240 g/m, and about 17 mm of preload in our installation.

Based on this prototype exoskeleton, we implement amount of weight-bearing and knee-assisting experiments, respectively (shown in Fig. 8). In the former test, we proved that the exoskeleton can carry up to 70 kg of weight, with high weight-bearing productivity. In knee-assisting torque measurements, we employed the tension dynamometer (Fig. 8b), to measure the amount of force used at different knee-joint leg flexion angles, with the arm of loading points remaining fixed to the rotation center of knee-joint. We then used high-definition camera to record both the measurement of dynamometer and the bending knee-joint with uniform speed. Lastly, through angle measurement software, to measure the forced applied in every 5° interval, and with 5 repeated measurements, we obtained the mean and standard deviation of experiment results.

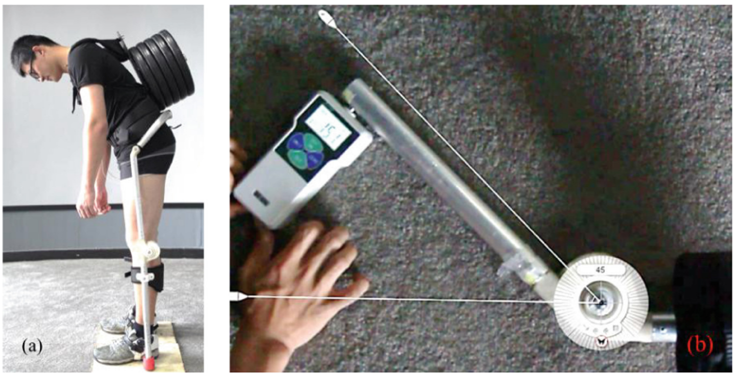

Fig. 8. Weight-bearing (a) and knee-assisting (b) experiments based on the prototype

In Fig. 9, the blue curve shows the analytical result of knee-assisting torque which is calculated according to Eq. 2, while the red curve indicates the mean value of experimental results with its error bars represent the maximum and minimum samples for interval bending angles. It can be observed that the analytical result matches the experimental result well. Overall, the assisting torque increases monotonically with the increase of the flexion angle. Specifically, the knee joint with eccentric contour exhibits multi-stage nonlinear assisting augmentation under different ranges of flexion angle. When $\theta \leq 50^{\circ}$, the torque increases slowest with θ, due to the short lever of string's stretch force; when $50^{\circ} \leq \theta \leq 80^{\circ}$, the torque rises up rapidly (steepest increasing slope among all range), mainly because the arm of force and deformation of spring expand simultaneously; when $\theta \geq 80^{\circ}$, the increasing momentum of the torque slows down again, since the arm of force does not keep increasing any more, but maintains around the maximum level about 50 mm.

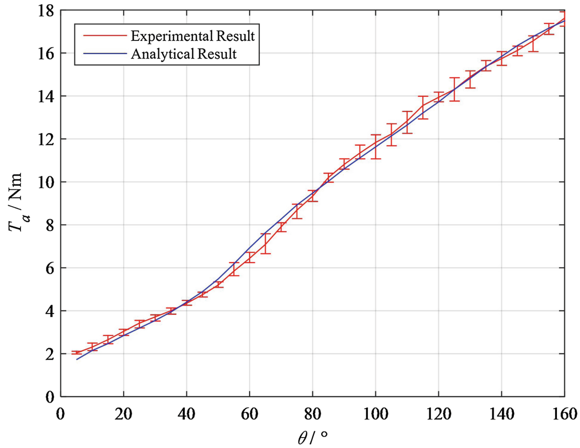

Fig. 9. The comparison between analytical and experiment results of the knee-assisting torque (Color figure online)

5 Conclusion

In our biomechanics research for human weight-bearing ambulation, we found that the knee joints flexion angle exhibits great differences between level walking and climbing. Normally, knee joint flexion angle does not exceed 60° during level walking, while knee joint flexion angle ranges from 65.8°–118.9° during climbing. With added weight on the walker, knee joint flexion angle increases even at the same altitude as when no extra weight was added. We decided to include pulley, springs, and strings in building a passive weight-assisting and energy saving knee-joint exoskeleton. Pulley has an eccentric contour, its curvature changes along with knee-joint flexion. Through theoretical analysis and calculations, we have derived deformation and force lever of the spring. Through experiment on the relation between knee-joint moment and flexion angle, we performed regression analysis on spring stiffness and preload. Results show that the actual spring stiffness is equivalent to the hypothesized value, and the analytical and experimental characteristics of knee-assisting moment match well. The device exhibits subtle assistance when flexion angle is less than 60°, while the assisting level gradually arises, which can grow up to 18 Nm, when the flexion angle further increases.

Nonetheless, we still face several limitations. Firstly, in the weight-bearing walking analysis, we tested for joint flexion angle, when we should also take into account thigh muscle activity during climbing. Another issue is the knee-joint's eccentric contour, in which we only have one applicable method so far, to obtain optimal assisting results, we need to think of better alternatives. Lastly, the mechanical verification experiment is not sufficient to prove that our passive knee-assisting device can reduce muscles' activities during weight-bearing ascending locomotion, and improve human metabolic productivity. Thus, an integrated experiment by simultaneously recording the muscle's EMG signal, metabolic energy consumption, and mechanical forces will be further pursued.

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References

1. Dollar, A.M., Herr, H.: Lower extremity exoskeletons and active orthoses: challenges and state-of-the-art. IEEE Trans. Robot. **24**(1), 144–158 (2008)
2. Slavka, V., Patrik, K., Marcel, J.: Wearable lower limb robotics: a review. Biocybern. Biomed. Eng. **33**(2), 96–105 (2013)
3. Mertz, L.: The next generation of exoskeletons: lighter, cheaper devices are in the works. IEEE Pulse **3**(4), 56–61 (2012)
4. Walsh, C.J., Pasch, K., Herr, H.: An autonomous, underactuated exoskeleton for load-carrying augmentation. In: 2006 IEEE/RSJ International Conference on Intelligent Robots and Systems, pp. 1410–1415 (2006)
5. Zoss, A.B., Kazerooni, H., Chu, A.: Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX). IEEE/ASME Trans. Mechatron. **11**(2), 128–138 (2006)
6. Justin, G., Ryan, S., Kazerooni, H.: Control and system identification for the Berkeley lower extremity exoskeleton (BLEEX). Adv. Robot. **20**(9), 989–1014 (2006)
7. Asbeck, A.T., Schmidt, K., Galiana, I., et al.: Multi-joint soft exosuit for gait assistance. In: 2015 IEEE International Conference on Robotics and Automation, Seattle, Washington (2015)
8. Wehner, M., Quinlivan, B., Aubin, P., et al.: A lightweight soft exosuit for gait assistance. In: IEEE International Conference on Robotics and Automation (2013)
9. Kawamoto, H., Lee, S., Kanbe, S., et al.: Power assit method for HAL3 using EMG based feedback controller. In: IEEE International Conference on Systems (2003)
10. Hercule. France (2017). http://www.rb3d.com/en/exo/
11. Young, A., Ferris, D.: State-of-the-art and future directions for robotic lower limb exoskeletons. IEEE Trans. Neural Syst. Rehabil. Eng. (2016)

12. Collins, S.H., Wiggin, M.B., Sawicki, G.S.: Reducing the energy cost of human walking using an unpowered exoskeleton. Nature **522**(7555), 212–215 (2015)
13. Wiggin, M.B., Sawicki, G.S., Collins, S.H.: An exoskeleton using controlled energy storage and release to aid ankle propulsion. In: 2011 IEEE International Conference on Rehabilitation Robotics (2011)
14. Kuo, A.D., Donelan, J.M., Ruina, A.: Energetic consequences of walking like an inverted pendulum: step-to-step transitions. Exerc. Sport Sci. Rev. **33**(2), 88–97 (2005)