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Design and Evaluation of a Novel Microprocessor-Controlled Prosthetic Knee

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ABSTRACT Transfemoral amputees demand a mechatronic lower limb prosthesis as technical substitute for restoring their gait functions. Prosthetic knee is the key component of a transfemoral prosthesis. The performance of the prosthetic knee determines the walking ability of the transfemoral amputee. This study proposed a novel microprocessor-controlled prosthetic knee with hydraulic damper and evaluated the performance of the prosthetic knee by function simulation and evaluation platform. The prosthetic knee with electrical-controlled hydraulic cylinder that could modulate knee flexion and extension damping properties independently and continuously by single motor was designed. Gait phase identification system based on knee angle sensor, inertial measurement units mounted on thigh connector and shank and force transducer embedded in shank was proposed. Gait phase identification and damping control strategy were determined by typical gait events during walking. Speed adaption and gait symmetry tests were conducted with a customized gait simulator to evaluate the performance of the proposed microprocessor-controlled prosthetic knee. The angle trajectories of the prosthetic knee were similar under a range of walking speeds. While the symmetry index values indicated that the stance phase was more asymmetry than swing phase, the peak swing flexion knee angles were consistently controlled between 60-70 degrees under different speeds. The knee angle symmetry was observed in different speeds during swing phase. It is suggested that the proposed microprocessor-controlled prosthetic knee could meet the fundamental demands of walking with smooth angular transition across different walking speeds.

INDEX TERMS Microprocessor-controlled prosthetic knee, hydraulic damper, gait phase identification, gait symmetry, peak swing flexion knee angle.

I. INTRODUCTION

There are more than 30 million amputees in the world [1], and lower limb loss in China is about 1.58 million [2]. The recovery of walking capacity in lower limb amputee is related to the level of the amputation. When the transfemoral amputation is completed, the patient needs a prosthetic leg to walk again [3]. In transfemoral amputees and patients with lower limb motor dysfunction, many have benefited from recent progress in advanced lower limb prosthetics and robotics [4]–[6]. Today, the basic needs of the transfemoral amputation for weight-bearing function during walking have been addressed.

However, the life quality of transfemoral amputee definitely has room for improvement. The key of such mobility improvement of transfemoral amputee is to develop prosthetic knee that meets higher functional demands, such as amputees' gait at self-selected walking speeds [7].

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The conventional prosthetic knee is a mechanical system which lacks the intelligent sensor and control system. Thus, conventional prosthetic knee cannot automatically adapt to the change of speed and locomotion modes, leading to abnormal and unsafe gait with increased energy consumption [8]. To deal with above issues, recent developments of prosthetic knees showed a trend towards the integration of microprocessor and bionic structure design to realize physiological gait for locomotion assistance.

Microprocessor-controlled prosthetic knees can be divided into two types: powered or active prosthetic knee and variable-damping or semi-active prosthetic knee [9]. Powered prosthetic knees have actuators that are able to provide extra energy to fulfil knee joint function during gait. The most common actuation system is an electrical drive with a transmission gear or a belt drive with ball screw transmission. Serial elastic elements, parallel spring and pneumatic actuated muscles are also used as actuators [10]. Lawson et al. presented the design and validation of a control system for a pair of powered knee and ankle prosthesis to be used as a prosthetic intervention for bilateral transfemoral amputees. The joints were actuated by brushless DC motors. Three stage transmissions consisting of a primary multi-V belt stage, a secondary roller chain stage, and a tertiary double-roller chain stage were adopted to drive the joints [11]. Budaker et al. presented the active knee driven by servomotor and bevel gear. The motor was combined with a bevel helical gearbox on the one side of the power train to simulate the right mechanical motion pattern. On the other side, a specially-designed motion controller helped to meet the implement of the targeted motion pattern [12]. Rouse et al. added a clutch in parallel with the motor within the series-elastic actuators to drive the powered prosthetic knee. The series elasticity was optimized to fit the spring-like torque angle relationship of early stance phase knee flexion and extension during level ground walking [13]. Although many powered prosthetic knees have claimed to provide functional benefit to user, the only powered prosthetic knee commercially-available to the end user is the Power Knee (Össur). The possibilities may be related to the reduced autonomy, lack of the silent and light-weighted actuators, as well as inadequate response when battery runs out, high cost and unnatural user interface [14].

In contrast, the variable-damping knee cannot provide active knee torque. The power source is the residual limb of the thigh. The damping is usually modulated by fluid control. Hydraulic, pneumatic and magnetorheological dampers are the most common choice [15]. Sensors are used to detect speed and gait phase during the gait cycle. Magnetorheological systems or valves are controlled by using this information to improve prosthetic performance. Solomon et al. designed a magnetorheological (MR) damper valve to control swing-phase damping in the transfemoral prosthesis. The results showed that the reduction by weight of the MR damper with designed valve was up to 71 % compared with the existing MR damper. However, the research only conducted the geometrical design variables optimization by mapping finite element analysis [16]. Richter et al. designed a virtual controller for a hydraulic prosthetic knee. The valve positions that minimized the difference between the actual moment and the moment demanded by the virtual controller were found. Although the simulation results showed that the knee angle tracking was possible with a root mean square average of 2.2 degrees, it failed to propose the valve and damper structure of the hydraulic prosthetic knee. The simulation was only conducted based on the concept design [17]. Recently, we have proposed a microprocessor-controlled prosthetic knee with hydraulic damper. The knee hydraulic system contained two separate needle valves to generate joint resistance for the flexion and the extension movement, respectively. The valves were controlled by linear motors. With the variation of valve opening, the flow resistance could be continuously varied from low to high values [18]. Typical commercialized variable-damping prosthetic knees are the Genium (Otto Bock), the C-Leg (Otto Bock), the SmartIP (Endolite) and the Rheo Knee (Össur) [19]. For the functional benefit and affordable weight and size, variable-damping knee has gained popularities in leading laboratories and industries.

The pneumatic damping is suitable for speed adaption in the swing phase, but is usually difficult to provide sufficient damping in the stance phase. The use of MR fluid has to avoid places with strong magnetic or electric field, which limits the application range [20], [21]. The hydraulic damper can provide strong damping with smaller size, which effectively ensure the stability of the stance phase. The turbulence property in the swing phase can also meet the requirements of flexibility.

The aim of this work was to propose a microprocessor-controlled prosthetic knee with a novel hydraulic damper and intelligent sense and control system. The electrical-controlled hydraulic cylinder that could continuously adjust knee flexion and extension damping properties by single motor was designed. Gait phase identification system based on knee angle sensor, inertial measurement units mounted on thigh connector and shank and force sensor embedded in shank was proposed. Gait phase identification and damping control strategy were determined by typical gait events. The gait simulation and evaluation platform of above-knee prosthesis was proposed. Speed adaption and gait symmetry tests were conducted to evaluate the performance of the proposed microprocessor-controlled prosthetic knee.

Compared to existing hydraulic damping prosthetic knees, the proposed microprocessor-controlled prosthetic knee has two main advantages. Firstly, only one motor was used in the electrical control for the hydraulic cylinder, while the knee flexion and extension damping adjustment were still independent and continuous. This structural design has potential to reduce the total weight of the prosthetic knee. Secondly, the gait phase identification system did not include special designed multidimensional torque sensor. All the sensors can be obtained easily from the market, which can facilitate the development of low-cost microprocessor-controlled

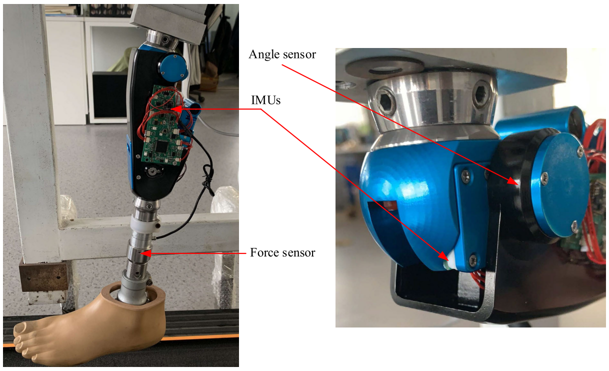

FIGURE 1. The prototype and the integrated sensor system.

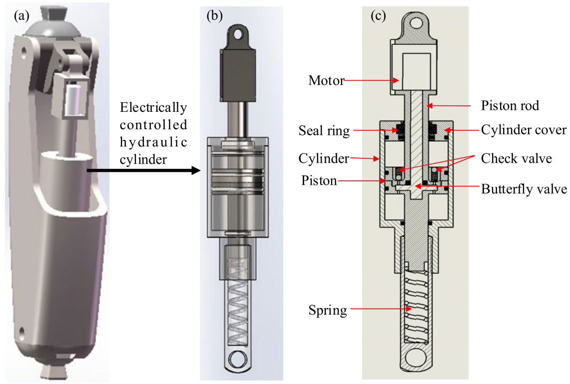

FIGURE 2. The mechanical structure of the microprocessor-controlled prosthetic knee.

prosthetic knee. The special ankle pylon also ensured that the gait phase identification system could avoid the effect of shoe change.

II. OVERVIEW OF THE MICROPROCESSOR-CONTROLLED PROSTHETIC KNEE

A. PROTOTYPE AND SENSOR SYSTEM

The prototype and sensor system are shown in Fig.1.The main targets of the sensor system of the prosthetic knee are the accurate recognition of gait phase and the detection of walking speed [22]. The walking process is cyclic and repetitive. The knee angle, lower limb posture and ground reaction force often directly reflect typical gait phases [23]. Gait phases are usually divided by typical gait events. Gait events can be determined using force or kinematic based measurement system, typically by means of footswitches such as force sensitive resistors (FSR) located in a shoe insole or gyroscope attached to the shank or prosthetic pylon [24]. The footswitches and gyroscope are suitable for obtaining on/off information. However, the in-shoe placement of the force sensors is not appropriate for changes of footwear. Another method is to use multi-axis force sensor such as six-axis force sensor. The commercialized microprocessor-controlled prosthetic knees such as Genium and RheoKnee both adopt multi-axis force sensor. However, multi-dimensional force sensors are expensive and this has become one of the major reasons of the high cost of the microprocessor-controlled prosthetic knees. To deal with above issue, a low-cost sensor system is proposed. Two miniature 9-axis inertial measurement units (LPMS-ME1, ALUBI) are used to obtain movement information of the thigh and shank. One is embedded into the upper linkage which is connected with the prosthetic socket, the other is placed in the circuit board which locates on the shank. LPMS-ME1 consists only of a 12-by-12mm multi-layer PCB to be integrated into our design as a surface-mounted component. An angle sensor (JA2236, EHSY) is mounted on the rotate axis of the knee to measure the knee angle. A load cell (JHBM-H3, HBM) measures the ground reaction force in the ankle pylon. The placement makes it that the change of artificial foot has no effect on the gait detection. The fabrication and assembly of the ankle pylon is easy. The motor (216000, Maxon) connects to a gear box with a 100:1 ratio drives the rotation of the adjusting valve.

B. DETAILED MECHANICAL STRUCTURE

The prosthetic knee (Fig. 2a) is a homotaxial joint design with an electrically controlled hydraulic damper (Fig. 2b). In the electrically controlled hydraulic damper (Fig. 2c), the upper push rod of the piston is pivotally connected to the upper leg segment of the prosthetic knee and the lower end of the cylinder is pivotally connected to the lower leg segment.

Therefore, in flexion the damper will contract and thus the piston will be driven downwardly in the cylinder by body load. In extension, the damper lengthens and the piston is pulled upwardly by body action. The microprocessor reacts at various transition points in the gait by activating a motor which adjusts the butterfly valve in the damper. The butterfly valve is able to vary flow port areas and fluid flow rates to thereby vary resistance to knee joint rotation in either flexion or extension at the same time. The artificial knee joint proposed by James et al. also employed a hydraulic damper with a single motor [25]. The main difference of this damper is the location of the flow orifices and the piston structure. The flow orifices of the damper in [25] are in the side wall of the piston, while the flow orifices in the proposed damper are at the bottom of the piston. The piston was divided into upper and lower part to form the hollow piston, while the piston in [25] was divided into inner and outer side. The location leads to the action force on the motor through the adjusting valve is very different. The balance between upper and lower oil pressure makes it easy to rotate the valve. In addition, the proposed damper structure is more easily manufactured.

The internal structure of the flow channel is shown in Fig.3. In the knee flexion, the piston is forced downwardly. The fluid in the lower chamber is pressed, fluid will flow upwardly through the check valve 1 and flow orifice 1. Fluid will not leave the lower chamber through the flow orifice 2 because the check valve 2 closes in this direction. In the knee extension, the piston is pulled upwardly. The fluid in the upper chamber are squeezed, fluid will flow downwardly through the check valve 2 and flow orifice 2. Fluid will not leave

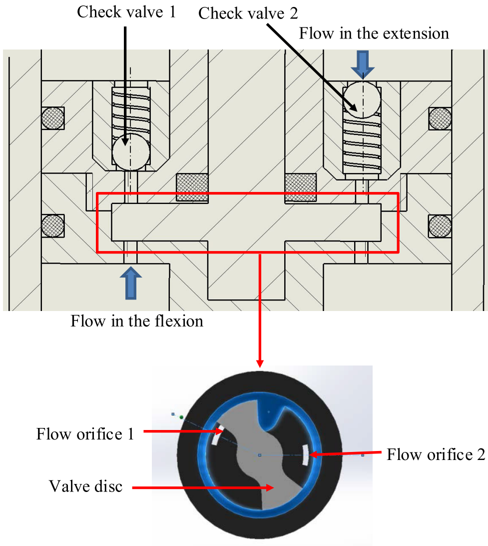

FIGURE 3. Internal structure of the flow channel and butterfly valve.

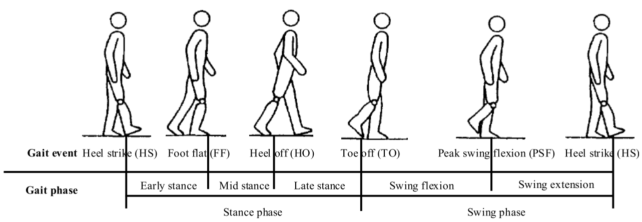

FIGURE 4. Human gait cycle.

the lower chamber through the flow orifice 1 because the check valve 1 closes in this direction. A steel spring stores energy during flexion and assists the subsequent extension movement. The diameter of the piston rod in upper and lower chamber is equal, so that there is no need to set a compensation reservoir.

III. IDENTIFICATION AND CONTROL STRATEGY

A. GAIT PHASE IDENTIFICATION

The human gait cycle can be divided into two phases: the stance phase and the swing phase. The detailed division of gait state is often obtained by typical gait events (Fig.4). The events and their resulting states are heel strike (HS) at early stance, then foot flat (FF) at mid stance, and heel off (HO) at late stance. The swing period is divided into swing flexion and swing extension. The swing flexion starts with toe off (TO) and ending with peak swing flexion (PSF) [26]. The swing extension starts with PSF and ending with full knee extension toward next HC.

The gait phases were represented in the form of a state machine with five distinct events. The transitions between the phases were governed by a knowledge-based algorithm. The inclined angle of the shank, direction of the angular velocity of the knee and the ground reaction force are combined to identify the gait phase during level walking. When the shank is perpendicular to the ground, the inclined angle is zero.

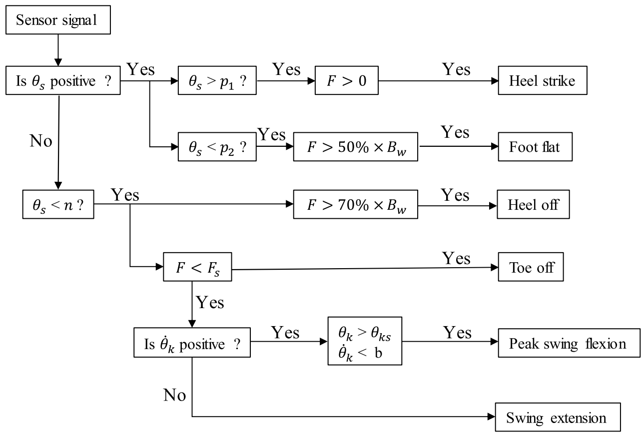

FIGURE 5. The principle of gait phase identification algorithm. θ_s: Inclined angle of the shank, p_1, p_2 : Preset positive values, n: Preset negative value, F: Force value, B_w: Body weight, F_s: Preset force value, $\dot{\theta}_k$: Knee angular velocity, θ_{ks}: Preset knee angle value, b: Preset knee angular velocity value.

The inclined angle is positive when the shank is in the front side. The inclined angle is negative when the shank is in the rear side. The positive value of knee angular velocity means swing flexion and negative value indicates swing extension. The inclined angle of the shank is obtained by the IMU on the shank. The gait phase identification principle is shown in Fig. 5. At the start of the gait cycle, as the heel makes contact with the ground, the inclined angle of the shank is positive and greater than the preset positive value, the force value is greater than zero. Thus, the heel strike event is detected as soon as above conditions are met. After the heel strike, the next event is foot flat, which begins when both the front and rear part of the foot touch the ground. This event is detected when the shank is nearly perpendicular to the ground and the ground reaction force is greater than fifty percent of the body weight. The inclined angle of the shank is positive and smaller than a preset value which is nearly zero. Then the subject lifts the heel and prepares to go into a swing phase, the heel off is detected when the inclined angle of the shank becomes negative and smaller than a preset negative value as well as the ground reaction force is greater than seventy percent of the body weight. When the inclined angle of the shank is negative and ground reaction force is nearly zero, it means the foot leaves the ground completely and swing flexion begins at toe off. The damper reduces the flexion of the knee until it reaches nearly zero angular velocity and the knee angle is greater than the preset value at peak swing flexion. After reaching peak flexion in swing, the knee begins to extend. The knee angular velocity becomes negative until it reaches the next heel strike.

B. DAMPING CONTROL STRATEGY

The corresponding damper behavior in typical gait events is described in Fig. 6. At the heel strike, the ground reaction force is high. The main function of the prosthetic knee is to absorb the terminal impact and support the body weight.

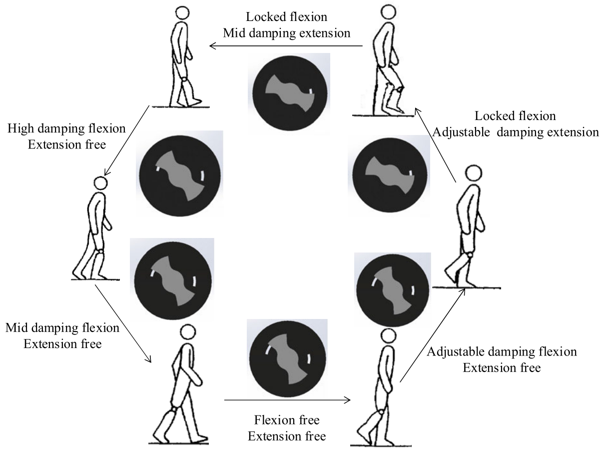

FIGURE 6. **Damping control strategy.**

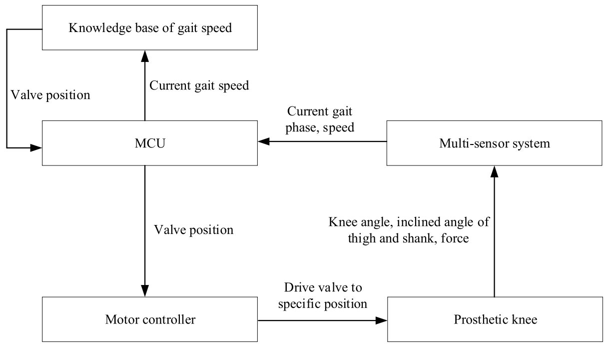

FIGURE 7. **Speed adaptive control flow block diagram.**

The prosthetic knee should be with high damping and stiffness. The flexion passage is closed via the rotation of the butterfly valve with the motor control. The knee flexion is locked and extension is in mid damping. During early stance, the flexion damping is high to allow the stance knee flexion and ensure the safety. The extension damping is set free in order to prepare for the stance extension. As the heel lifts off the ground, while the forefoot is still on the ground, the knee angle is reduced as the knee starts extending again. The flexion damping is still high and extension damping is moderate to reduce the extension impact. Swing flexion begins at toe off. The butterfly valve is in the position that the flexion and extension passage both open completely. The flexion and extension damping are both free to allow for swing flexion. As the knee flexes beyond specific degrees, the adjustable flexion damping control is implemented to reduce hyper-flexion of the knee until it reaches zero velocity at peak swing flexion angle.

The speed adaption of the microprocessor-controlled prosthetic knee mainly reflects in the control of peak swing flexion. It means that the flexion damping from the knee in swing phase is proportional to the walking speed. When the knee moves faster, higher flexion damping force is required to slow it. Adjustable flexion damping force is the main advantage over passive mechanical knees which the damping level cannot be changed automatically during walking. Speed identification is the premise of speed adaptive control strategy. Several methods are usually used to approximate walking speed. One direct method is to measure the stride time, longer stride time indicates slower walking speed. However, the stride time is always known after the stride is completed. It means that the walking speed obtained by stride time is always one step old. To overcome the obstacle, the contact time of the stance phase before a given swing phase is an alternative. Previous study finds that the contact time (time from heel strike to toe off) and walking speed have a strong inverse correlation [27]. Thus, the contact time of all the speeds are stored. The valve position is controlled by comparing the differential of contact time for the stance phase in the sequential gait cycle. When the absolute value of time error is less than the set value, the walking speed is thought as no change, the valve position keeps same with previous step. When the time error is greater than the set value, the walking speed is thought as decrease compared to forward step, the open size of the valve increases to reduce the damping force. When the time error is smaller than the negative set value, the walking speed is thought as increase compared to forward step, the open size of the valve decreases to increase the damping force. The target peak swing flexion angle is set as sixty degrees. The speed of damping adjustment is proportional to the error between the target angle and actual peak swing flexion angle measured by the angle sensor. After reaching maximum flexion in swing, the knee begins to extend, and the extension damping is controlled as an adjustable damper to decelerate smoothly the motion of the swinging leg in preparation for the heel strike of the subsequent gait cycle. The speed adaptive control flow block diagram is shown in Fig.7.

IV. EXPERIMENTS AND RESULTS

A. GAIT SIMULATION AND EVALUATION PLATFORM

The traditional evaluation criterion for lower limb prostheses mostly depend on the subjective feelings of the amputees instead of quantitative performance. Subjective feeling tests are poor for insurance, time consuming and tedious. There are liability and safety concerns due to the risk of falling or stumbling. Robotic testing can facilitate the development of new concepts, designs and control systems for lower limb prostheses. In order to make quantitative evaluation of the microprocessor-controlled prosthetic knee, the gait simulation and evaluation platform of above-knee prosthesis is designed. The principle and prototype of the platform are shown in Fig.8.

The mechanic frame of the proposed simulator consists of intact leg modular and prosthetic leg modular. The prototype is self-made. The designed prototype applies treadmill to simulate level walking of the intact and prosthetic leg. Lifting air cylinders are employed to simulate the gravity shift of human body. Hip and knee joint of intact leg are driven by motors. Hip joint of prosthetic leg is driven by motor. Microprocessor-controlled prosthetic knee is connected with the simulator by

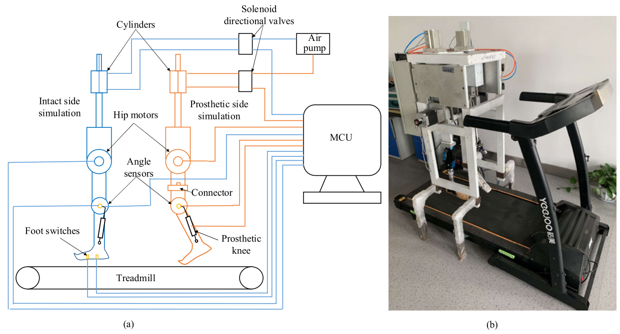

FIGURE 8. **The principle and prototype of the platform. (a) Design principle, (b) Prototype.**

a special joint to achieve the function simulation. Normal hip angle data is inputted to driven motors of intact and prosthetic leg. Normal knee angle data is inputted to intact simulation leg. The simulator has two major functions: (1) Simulation of normal hip and knee angle under different speeds and the real-time detection of angle; (2) Simulation of the hip drive of the prosthetic leg to evaluate the knee joint performance. Knee angles of intact and prosthetic leg are recorded by angle sensors.

B. SPEED ADAPTION, GAIT PHASE IDENTIFICATION AND GAIT SYMMETRY TESTS

The proposed simulation and evaluation platform of microprocessor-controlled prosthetic knee was used to conduct the speed adaption, gait phase identification and gait symmetry tests. The trend of different people during each gait phase are similar. The hip and knee angles are only used to verify the tracking performance of the controller. What we need are walking data of a complete gait cycle. The data obtained by the means of many sequential gait cycles. The diversity of curves has little effect on the tracking performance. It is sufficient to obtain data from one healthy person. The subject was required to walk on the treadmill in three specific velocities, i.e., 0.6m/s, 1.1m/s and 1.6m/s. The kinematic parameters were measured through a real-time 3D gait and motion analysis system (RealGait, JIANGSU NEUCOGNIC MEDICAL COMPANY, LTD. China). This system consists of seventeen inertial sensors. Data collection and analysis of the whole body were achieved synchronously. The hip and knee joint trajectories of the healthy subject under three speeds are shown in Fig.9. The data presented here on hip and knee angles are restricted to motion in the sagittal plane. These data served as the input of the following function simulation and evaluation platform.

Normal gait is a cyclic and symmetric process, depending on the continuous and complex transfer of energy among the body segments. This allows mobility with minimum energy expenditure. With the increase of the asymmetry of the cycle, the energy expenditure increases [28]. Symmetry appears to be a relevant aspect for differentiating a normal and pathological gait. Lower limb amputees experience

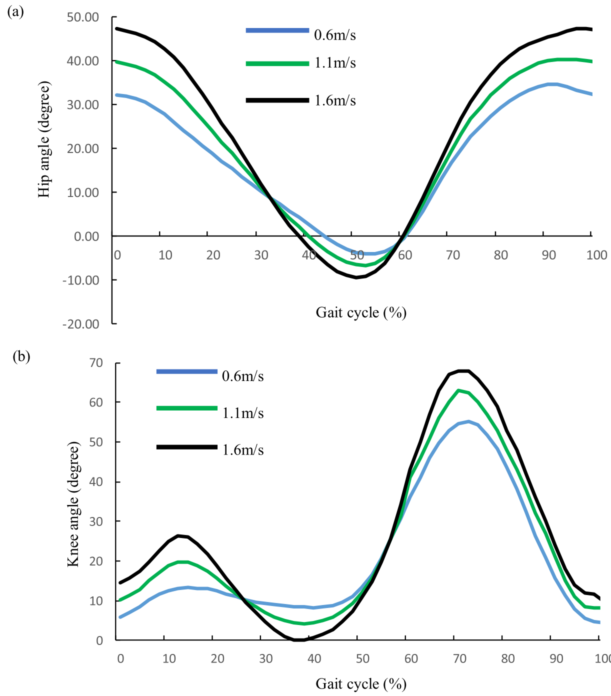

FIGURE 9. **(a) Hip angle during one cycle under different speeds, (b) Knee angle during one cycle under different speeds.**

exhibits pathological gaits which are often asymmetric and less stable than non-amputees. The gait symmetry compares the kinematics of the healthy leg and prosthetic leg. The gait symmetry represents the adaptability to speed changes of the microprocessor-controlled prosthetic knee. It is a key index of the performance of the prosthetic knee [29]. The gait symmetry quantification used three indices introduced by Karaharju-Huisman *et al.* [30].

(1) Symmetry Index (SI)

$$SI = \frac{2(X_R - X_L)}{X_R + X_L} \times 100\%$$

where X_R is the right limb data and X_L is the corresponding data for the left side. The prosthetic leg side was X_L and healthy side was X_R in this work. SI $= 0$ represents perfect symmetry and $+/-$ showing limb dominance.

(2) Ratio I

$$R_I = \frac{X_R}{X_L}$$

where $R_I = 1$ represents perfect symmetry.

(3) Ratio II

$$R_{II} = \frac{(X_R - X_L)}{\max(X_R, X_L)}$$

where $R_{II} = 0$ represents perfect symmetry and $+/-$ showing limb dominance.

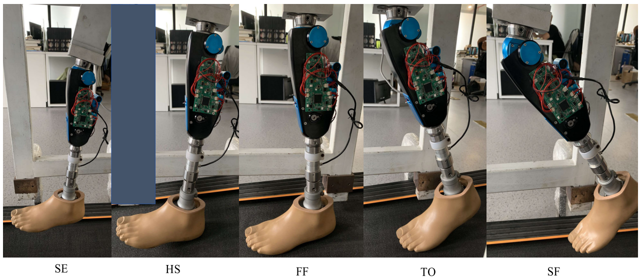

FIGURE 10. **The typical gait events and phases during test. SE: Swing extension, SF: Swing flexion.**

TABLE 1. **Gait phase identification results.**

Speed (m/s)	Gait phase	Test cycles	Ident. cycles	Recog. rate (%)	Total recog. rate (%)
	ES		50	100	
	MS		49	98	
0.6	LS	50	48	96	98.8
	SF		50	100	
	SE		50	100	
	ES		50	100	
	MS		48	96	
1.1	LS	50	48	96	98
	SF		49	98	
	SE		50	100	
	ES		50	100	
	MS		47	94	
1.6	LS	50	46	92	96.4
	SF		48	96	
	SE		50	100	

C. RESULTS

Stance and swing phase were detected by the proposed sensor system. The typical gait events and phases during test are shown as Fig.10.

Fifty gait cycles during stable state were chosen to determine accuracy of the gait phase identification under each speed. The results of gait phase identification accuracy were shown in Table 1. The walking speeds play a greater influence on gait phase recognition rate of stance phase. The recognition rate of mid stance decreases with the increase of speed. The major reason is that the stance phase becomes shorter with higher speed. The speeds have less effect on the recognition rate of swing phase. The gait phase recognition rate has little significance of the whole gait phases except mid stance. While the total recognition rate decreases with speed increase, the total recognition rate is 96.4% in the highest speed (1.6m/s). The lowest total recognition rate is still enough for the safe walking. It is suggested that the gait phase identification accuracy can meet fundamental walking speed demands.

Forty-five gait cycles with a hundred percent accuracy of gait phase recognition rate were chosen to analyze the symmetry indices between prosthetic and intact knee angle under different speeds. The angles of intact and prosthetic knee in the same phase were equilibrated and translated into percent of gait to show the repeatability.

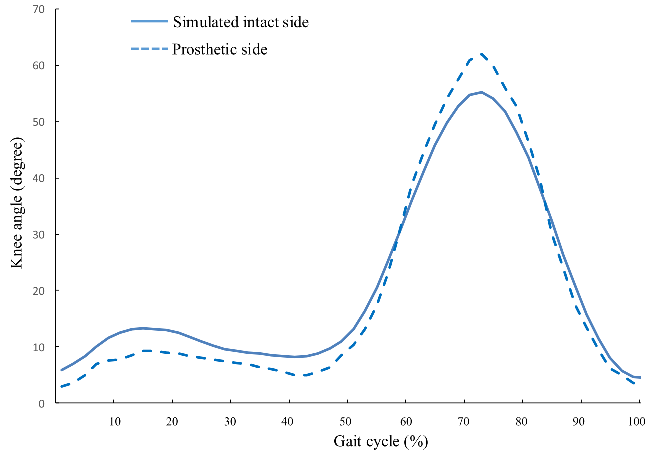

FIGURE 11. **The knee angle of the both sides in 0.6 m/s.**

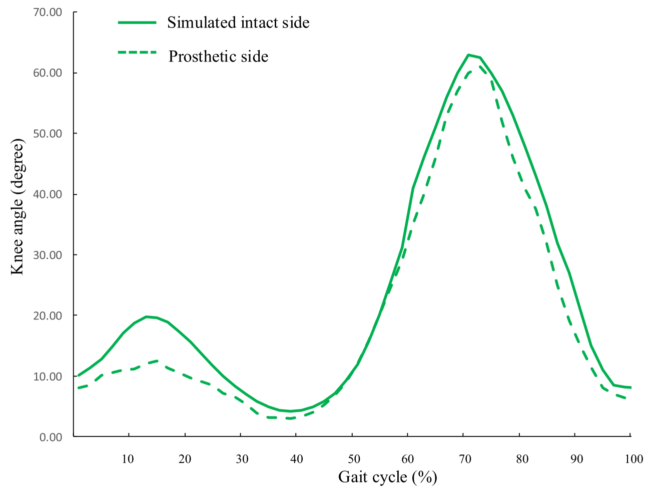

FIGURE 12. **The knee angle of the both sides in 1.1 m/s.**

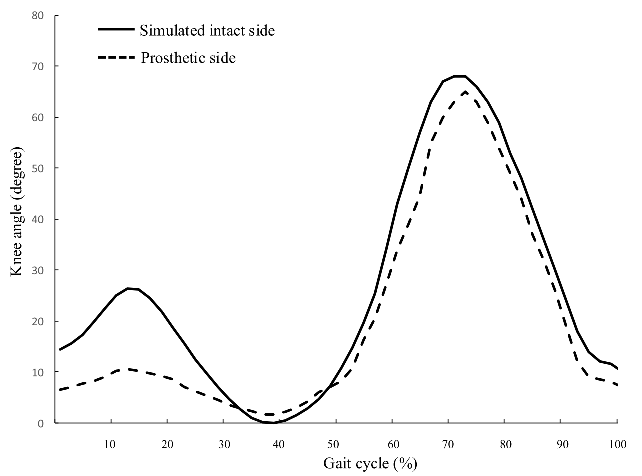

FIGURE 13. **The knee angle of the both sides in 1.6 m/s.**

The intact and prosthetic knee angle trajectories for slow walking speed (0.6 m/s), normal walking speed (1.1 m/s) and fast waking speed (1.6 m/s) were shown in Fig.11, Fig.12 and Fig.13. The symmetry indices data calculated under different

TABLE 2. Symmetry indices data calculated under different speeds.

Speed (m/s)	Gait phase	SI	R_I	R_{II}
0.6	Stance	25.3%	1.29	2.29
	Swing	-4.6%	0.95	-0.50
1.1	Stance	24.5%	1.28	2.25
	Swing	11.8%	1.13	1.36
1.6	Stance	24.7%	1.67	3.16
	Swing	13.1%	1.14	1.55

walking speeds were shown in Table 2. The knee angle trajectories of prosthetic side were similar under different speeds. The peak swing flexion knee angles were controlled between sixty and seventy degrees under different speeds. However, the knee angle symmetry differed by gait phase. There was a significant difference in symmetry between stance and swing. The symmetry index values indicated that the stance phase was more asymmetry than swing phase. This is due to the stance pre-flexion was smaller than the normal gait. The stance phase kinematic symmetry, in most cases, depended on differences in the knee position during early stance and mid stance. In the swing phase, the symmetrical kinematics was observed in different speeds. The number values of SI were smaller than 15% in swing phase.

V. DISCUSSION

The microprocessor-controlled prosthetic knee with a novel hydraulic damper and intelligent control system was proposed. The electrical-controlled hydraulic cylinder which adjusted knee flexion and extension damping independently and continuously by single motor was designed. The weight and energy consumption of the microprocessor-controlled prosthetic knee were reduced due to the single motor design. Compared to previous work [18] which two linear motors are used to control two needle valves separately, the butterfly shaped valve and single motor lead to smaller size and longer work time. Without the ability to generate power within the knee, the proposed prosthetic knee is still an energetically passive device. However, the knee joint is equivalent to semi-active joint in normal human walking [31]. The positive power is necessary for ascending inclines and stairs. The passive devices could not create as much flexion at toe off as a biological knee for sufficient heel rise to avoid hip hike. Fully powered knee prosthesis is often loud, expensive, and has a limited battery life. Thus, instead of relying exclusively on passive damping or fully powered movement, semi-active prosthetic knee takes advantage of the benefits offered by both types of knees is the important direction of the prosthetic knee design. Park et al. proposed a prosthetic knee equipped with the electronically commutated (EC) motor and MR damper. The active torque was provided by the EC motor and the damping torque was obtained from the MR damper. While the results demonstrated that the desired knee joint angle was achieved in different walking velocities on the ground, the problem that how to obtain acceptable motor size and energy consumption under providing enough active torque was still difficult to solve [32]. Awad et al. developed a semi-active prosthetic knee, which could work in both active and passive modes based on the energy required during the gait cycle of various activities of daily livings. This prosthetic knee had back-drive capability to operate passively in unactuated phase depending on the amputee-prosthesis-environment system dynamics in addition to providing assistive power in actuated phase when positive energy was required [33]. Geeroms et al. proposed a novel semi-active actuator with a lockable parallel spring for a prosthetic knee joint. The proposed novel actuator reduced the energy consumption for the same trajectory with respect to a compliant or directly-driven prosthetic active knee joint and improved the approximation of healthy knee behavior during level walking compared to passive or variable damping knee prosthesis [34].

Our results also showed that the peak swing flexion knee angles were controlled between sixty and seventy degrees under different speeds. The peak prosthetic knee flexion did not increase with speed. The result is consistent with the work of Herr and Wilkenfeld [35]. They presented a magnetorheological knee prosthesis that automatically adapted knee damping to the gait of the amputee using only local sensing of knee force, torque, and position. The user-adaptive knee gave a maximum flexion angle that always was less than 70 degrees and agreed well with the unimpaired, biological data. Lura et al. also determined differences between the knee flexion angle of persons using the Genium knee and the C-Leg knee. For over ground walking, Genium knee use resulted in an average swing phase peak knee flexion increase of 5-7 degrees for all walking speeds except during fast walking, which resulted in two degrees increase that was not significantly different from C-Leg. The result was also similar with this work [36]. Previous work [18] also conformed the importance of the maximum swing flexion knee angle for speed adaptive control. The relevant results indicated that the proposed microprocessor-controlled prosthetic knee in this work had good performance in the speed adaptive control.

The gait symmetry results showed that the knee angle symmetry varies during different gait phases. There was a significant difference in symmetry between stance and swing. The symmetry index values indicated that the stance phase was more asymmetrical than swing phase. The result was in agreement with the work of Kaufman et al. [37]. Our results demonstrated that there was a significant difference in gait symmetry between stance and swing with the greatest asymmetry occurred during stance. The stance phase kinematic asymmetry, in most cases, depended on differences in the knee position during loading response.

This study had several limitations. Firstly, only the knee angle symmetry was analyzed in the current work. The gait of amputees has asymmetry in the temporal parameters, ground reaction forces, and center of pressure trajectories. Many other biomechanical factors should be compared in the following work. Secondly, the human tests of the transfemoral amputees wearing current prosthetic knee should be conducted in the future research. Finally, the simulation of the level walking was not sufficient for representing the real road conditions. Different terrain tests should be performed in the future.

VI. CONCLUSION

The work proposed a microprocessor-controlled prosthetic knee with a novel hydraulic damper and intelligent sense and control system. The homotaxial knee joint with electrical-controlled hydraulic cylinder which adjusted knee flexion and extension damping independently and continuously by single motor was designed. Gait phase identification system based on knee angle sensor, inertial measurement units placed on thigh connector and shank and force sensor embedded in shank was proposed. Gait phase identification and damping control strategy were determined by typical gait events. The gait simulation and evaluation platform of above-knee prosthesis is proposed. Speed adaption and gait symmetry tests were conducted to evaluate the performance of the proposed microprocessor-controlled prosthetic knee. The knee angle trajectories of prosthetic knee were similar under different speeds. The peak swing flexion knee angles were controlled between sixty and seventy degrees under different speeds. The symmetry index results indicated that the symmetry was acceptable. The proposed microprocessor-controlled prosthetic knee could meet the demands of walking.

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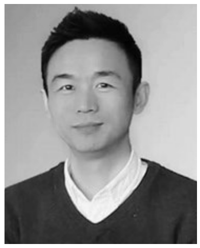

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