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To Calculate Reference Movement of Impaired Knee in Real-time Based on Biomechanical Information of the Contra-lateral Healthy Limb

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Abstract: Reference movement provided for an exoskeleton prosthesis that assists patients in walking is usually given based on statistical gait data of healthy individuals. However, it is easy to cause misidentification of gait's phase, discontinuity of reference movement and serious asymmetry of two limbs. This study proposes a method to give a reference movement for impaired knee based on biomechanical information of the contra-lateral healthy limb. This can enhance movement symmetry of two limbs. Different phases are identified according to the statuses (peak, valley and zero values, increasing or decreasing) of knee angle, angular velocity, forefoot reaction force and heel reaction force of the healthy limb. The specific statuses rather than numerical values are favourable to enhance accuracy of phase identification and continuity of reference movement. Sine or cosine function is used to express motion trajectory of knee approximately and the period and peak value of the output function can be regulated according to the measured knee angles of the healthy limb. Thus, the reference movement can match that of the healthy knee in real-time even though walking pace and stride length are changed. Verification experiment was performed on four subjects and the results confirmed the feasibility of the proposed method.

Key words: Gait analysis, adjustable walking pace and stride length, real-time response, symmetry movement, reference movement

INTRODUCTION

Recently, the number of disabled patients has been increasing rapidly because of traffic accident, industrial accident and disease. Moreover, the number of elderly patients with motor dysfunction is increasing distinctly as the transformation of social structure. In addition, many survivors lost limbs because of natural disasters such as the earthquakes broke out in Wenchuan anf yushu in Sichuan¡Province. To provide walking auxiliary equipment for these disable patients is a main means to help them achieve independent walking and return to the mainstream of society again. In order to control a walking auxiliary device to achieve a natural gait, accuracy of its reference movement is very important (Wang and Jin, 2002) (Gao *et al*., 2011). However, in the stage of providing reference movement for an above-knee artificial limb or exoskeleton prosthesis, there still exist the following issues:

- To identify gait phases in view of some numerical values of statistical gait data of healthy individuals (Pappas *et al*., 2001) (Kirkwood *et al*., 1989) (Sabatini *et al*., 2005). However, as for different persons or one person in the case that he changes stride length or walking pace, there will be a clear difference in gait parameters (Han and Wang, 2011; Di *et al*., 2012). Therefore, it is easy to cause misidentification of a phase and discontinuity of calculated reference movement. This is unfavorable to control the stability of an artificial limb or exoskeleton prosthesis and further, the safety of patients cannot be ensured.
- To provide a reference movement for an artificial limb or exoskeleton prosthesis by applying statistical gait data of healthy individuals (Zeng *et al*., 2013; Riener *et al*., 2005). However, there is an obvious difference among individual movements. If an exoskeleton prosthesis that directly applies statistical gait as reference movement is used to assist different patients, worse movement symmetry of two limbs will be obtained. This is not preferable for an artificial limb to assist patients to walk steady and it is easy to cause discomfort of patients
- It is difficult to regulate walking pace and stride length of the reference movement by using statistical gait data (Zlatnik, 1999; (Herr and Wilkenfeld, 2003; Gaschler, 2011). Thus, an artificial limb or exoskeleton prosthesis cannot support patients to walk freely and comfortably in various external environments

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In view of the above research background, this study proposes a new method to give a reference movement for impaired knee in real time based on the movement of the healthy knee. By referring to the movement of patients' own healthy knee, the accuracy of phase identification, the continuity as well as the adjustability of stride length and walking pace of the calculated reference movement can be enhanced.

EXPERIMENTAL SECTION

Reference movement of impaired knee: Definition of rotational angle and the corresponding directions of hip, knee and ankle are presented in Fig. 1, in which O_1, O_2 and O_3 are the corresponding rotational center, θ_{hip}, θ_{knee} and θ_{ankle} denote the three joint angles. Movement of knee in the sagittal plane is mainly studied in this study.

One gait cycle is divided into eight phases, including foot flat, earlier-stance, mid-stance, later-stance, pre-swing, earlier-swing, mid-swing and later-swing (Fig. 2). In this study, the corresponding relationship of two lower limbs' phases (in the following context, limb denotes lower limb) are defined as: The foot flat, earlier-stance, mid-stance and later-stance phases of the right limb are corresponding to the pre-swing, earlier-swing, mid-swing and later-swing phases of the left limb respectively and the pre-swing, earlier-swing, mid-swing and later-swing phases of the right limb are corresponding to the foot flat, earlier-stance, mid-stance and later-stance phases of the left limb respectively. Therefore, as for a patient with one limb impaired, the reference movement of the impaired limb can be given based on the corresponding movement of the healthy limb in the first half of the gait cycle. For example, if the current phase of the healthy limb is foot flat, the reference output of the impaired limb can be given based on corresponding movement of the healthy limb in the pre-swing phase in the first half of the gait cycle.

During walking, stride length and walking pace may be changed in any time. Accordingly, the two phases of two limbs with an interval of half a gait cycle may be different. In order to give a relatively accurate reference movement for impaired knee, even though there is a change in stride length or walking pace, motion trajectories of knee in the eight phases are approximated by a series of sine or cosine functions. In each phase, cycle time (T), start angle and finish angle of the healthy knee are recorded. After half a gait cycle, 4 times of T and the difference between the finish angle and start angle are respectively used as the period and peak value (A) of the approximate output function (give a reference movement for the impaired/contra-lateral knee) of the current phase (Table 1). Whereas in the phase of later-swing, the duration between the start time and the time at the turning point is considered as T and the angle difference between

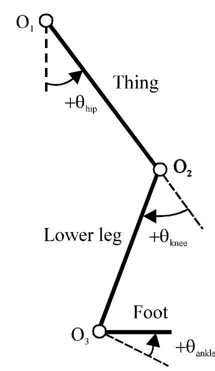

Fig. 1: Definition of joint angles and rotational directions of lower limb

Table 1: Approximate functions expressing the motion trajectories of two knees in different phases

Phase cycle	Ipsilateral: Healthy limb; Contra-lateral: impaired limb	knee angle functions of the two limbs, t: current time
Foot flat	Ipsilateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
	Contra-lateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
Earlier-stance	Ipsilateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
	Contra-lateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
Mid-stance	Ipsilateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
	Contra-lateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
Later-stance	Ipsilateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
	Contra-lateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
Pre-swing	Ipsilateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
	Contra-lateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
Earlier-swing	Ipsilateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
	Contra-lateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
Mid-swing	Ipsilateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
	Contra-lateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
Later-swing	Ipsilateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
	Contra-lateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A
Turning point of later-swing	Ipsilateral	A*sin(3.14*(t-start_t)/(2*T))+phase_start_angle
	Contra-lateral	-A*cos(3.14*(t-start_t)/(2*T))+phase_start_angle+A

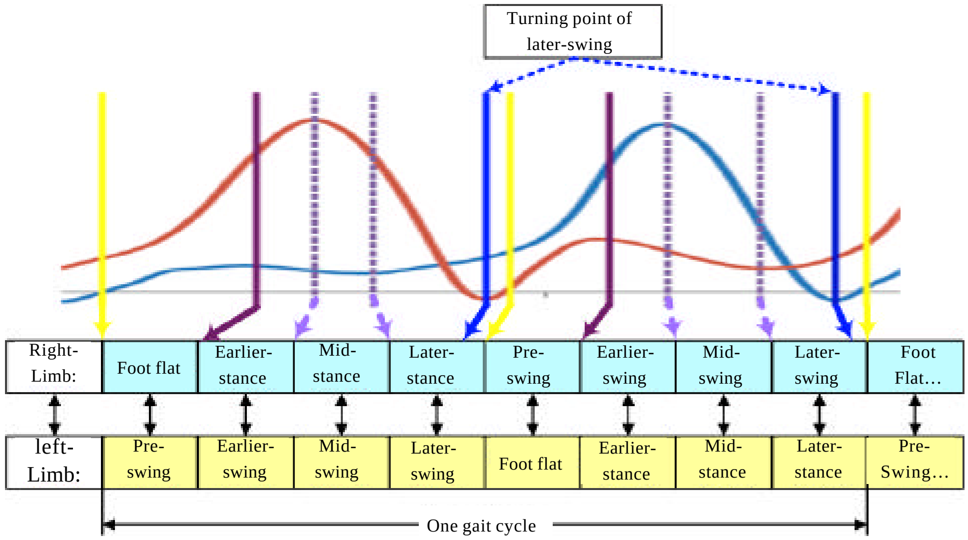

Fig. 2: Representative motion trajectories of two knees in one gait cycle (blue: right limb; red: left limb) and the corresponding phase relationship

Table 2: Corresponding parameters of approximate functions in the previous 4 phases in the 1st step

Phase cycle	Corresponding parameters of the knee angle functions
Foot flat	A = 20, T = 0.12, phase_start_angle = 0;
Earlier-stance	A = 15, T = 0.1, phase_start_angle = 20;
Mid-stance	A =-20, T = 0.2, phase_start_angle = 35;
Later-stance	A =-15, T = 0.12, phase_start_angle = 15;
Pre-swing	A = 10, T = 0.12, phase_start_angle = 0;
Earlier-swing	A =-4, T = 0.1, phase_start_angle = 10;
Mid-swing	A =-4, T = 0.2, phase_start_angle = 6;
Later-swing	A = 6, T = 0.18, phase_start_angle = 2;
Turning point of later-swing	A = 6, T = 0.18, phase_start_angle = 2;

Since the 1st phase in the 1st step is foot flat or pre-swing, the corresponding phase start angle in these two phases equal to zero; besides, if the device only supports the healthy limb to start swing, the parameters in 1st~4th phase cycles are useless

these two time points is considered as A. By using the above method, the period and A of the approximate output function of the current phase can be regulated accordingly, so as to make the reference movement adapt to the real-time change in stride length or walking pace.

However, in the first four phases of the first step, the reference movement of the impaired limb cannot be given based on the corresponding movement of the healthy limb in the first half of the gait cycle. Therefore, the corresponding period and A of the approximate output functions are defined based on statistical gait data of healthy individuals. The applied parameters are listed in Table 2. In our proposed method, only in the first half of the first gait cycle the reference movement of the impaired limb is given based on statistical gait data, the continuity of calculated reference movement will not be affected. Moreover, impaired patients usually start walking with the healthy limb moving forward first. Accordingly, the first four phases of the impaired limb are foot flat, earlier-stance, mid-stance and later-stance, that is, the corresponding reference movement of the impaired knee have small angles. Thus, the stability of the two lower limbs will not be affected too.

Phase identification: Characteristics of knee angle, angular velocity and reaction forces of forefoot and heel are used for identifying gait phases, refer to Fig. 3 and Table 3. In order to avoid inaccurate identification caused by individual differences or by changes in stride length or walking pace, statuses of peak value, valley value, zero value, increasing or decreasing are considered as the main characteristics. In the program, based on the status of current phase, the main characteristics of the above 4 variables are applied only for identifying the following phase, as shown in the flow chart (Fig. 4) and the explanation of defined variables is given in Table 4. This is favorable to achieve higher efficiency and accuracy of phase identification.

When a new phase is identified, the following tasks will be performed at the start time of the current phase:

- Refresh the status of 'input_period'
- Refresh the information (cycle time and finish angle) of the previous phase based on the start time and the

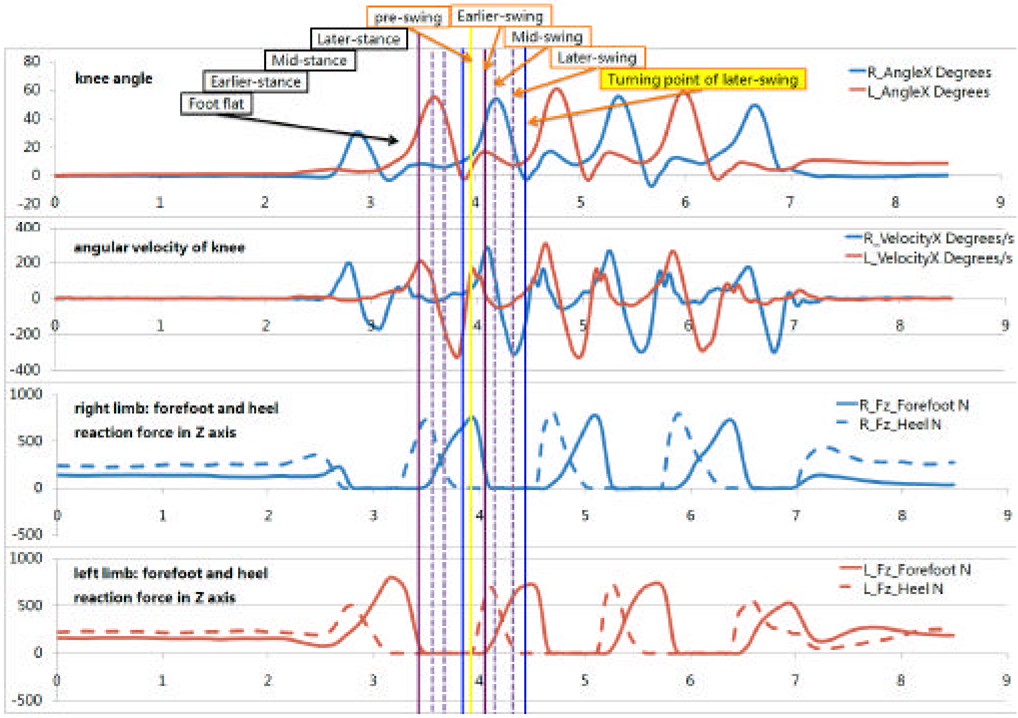

Fig. 3: Knee angles, angular velocities, forefoot reaction forces and heel reaction forces of the two limbs (blue: Right limb; red: left limb)

Table 3: Conditions used for identifying gait phases

Phase cycle	Knee angle	Angular velocity of knee	Forefoot reaction force	Heel reaction force
Foot flat	Increase	Positive peak		Increase
Earlier-stance	Positive peak		Increase;>5; less than heel reaction force	
Mid-stance	decrease	Negative peak	Increase	decrease
Later-stance	Negative peak		Increase; greater than heel reaction force	decrease
Pre-swing	Increase	Increase	Positive peak	
Earlier-swing	Increase	Positive peak	decrease	< 5
Mid-swing	positive peak		< 5	< 5
Later-swing	decrease	Negative peak	< 5	< 5
Turning point of later-swing	Negative peak	---	---	---

Table 4: Meanings of the main variables

Variables	Meaning
First_step	0: the steps except for the 1st one; 1: the 1st step;
Last_step	0: the steps except for the last one; 1: the last step;
Count	0: phase statuses in the steps except for the 1st one; 1~9: The 1st~9th phase status in the 1st step (including the turning point of Later-swing)
Input_period	0: no movement; 1~9: Corresponding to foot flat, Earlier-stance, Mid-stance, Later-stance, pre-swing, Earlier-swing, Mid-swing, Later-swing and the turning point of Later-swing.
T	Phase cycle of the current phase.
A	Peak of the output function corresponding to the current phase.
Start_time	Start time of the current phase.
Phase_start_angle	Start angle of the output function corresponding to the current phase.
Limb_info(3)	Structure array, the three elements are the corresponding data (knee joint information and foot reaction force) of the healthy limb obtained in three consecutive sampling periods.
Period_info(9)	Structure array, the 9 elements are the phase status information (phase cycle, start angle and finish angle) corresponding to 9 phase periods
Phase_StartPoint_Num	The sampling point number start from the start point of the current phase.
interval_point_Num	Interval sampling point number, the current phase cycle will be re-estimated every interval of this number.
alpha1_delta	The difference between knee angle every interval of 'interval_point_ Num' and the start angle of the current phase
alpha2_delta	The difference between knee angle every interval of '2*interval_point_ Num' and the start angle of the current phase

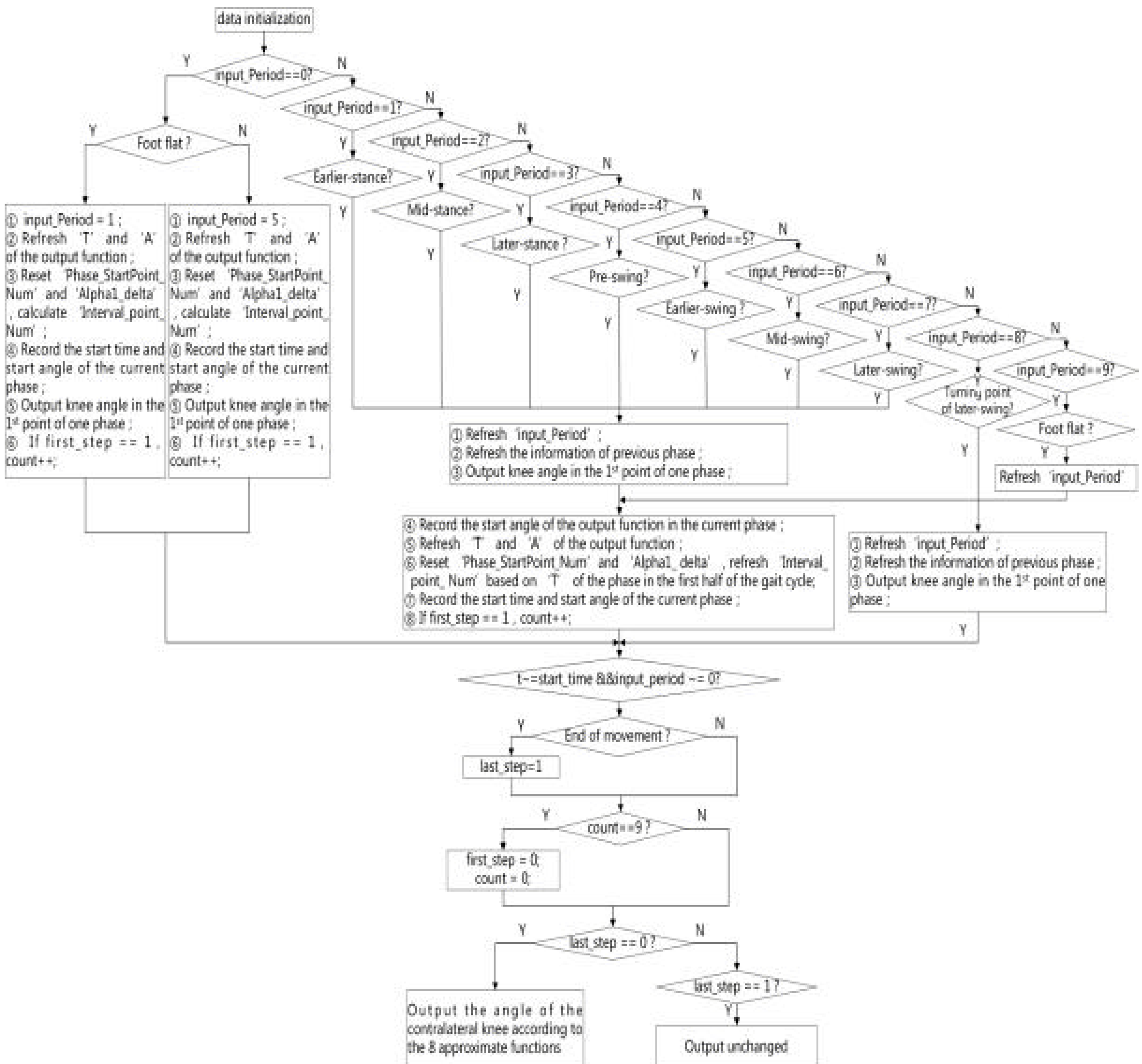

Fig. 4: Software flow chart

start angle of the current phase. (as for the healthy/ipsilateral limb)

- Output the reference angle for the impaired knee. (as for the impaired/contra-lateral limb)
- Refresh the 'phase_start_angle' of the approximate output function of the current phase with the value achieved in step 3. (as for the impaired/contra-lateral limb)
- Refresh T and A of the approximate output function (as for the impaired/contra-lateral limb) of the current phase base on the cycle time, start angle and finish angle of the corresponding phase in the first half of the gait cycle (as for the healthy/ipsilateral limb)
- Reset 'phase_StartPoint_Num' and 'alpha1_ delta' and calculate 'interval_point_Num' based on the T (as for the healthy/ipsilateral limb). 'interval_point_Num' is used for regulating T of the current phase after interval_point_Num sampling points and further to make the reference movement adapt to the change in stride length or pace
- Record the start time (start_time) and the start angle (the second member of the corresponding structure

elements of period_info(9)) of the current phase (as for the healthy/ipsilateral limb)

- If 'first_step' equals to one, 'count' pluses one

In the first phase of the first step, T and A of the current phase and interval_point_Num are calculated based on the statistical gait data of healthy individuals rather than the information of the corresponding phase in the first half of the gait cycle and the reference angle at the start time for the impaired knee is considered as zero. Additionally, in the turning point of later-swing, only the tasks 1-3 are performed. Accordingly, in the following phase of foot flat, only the tasks 1, 4, 5, 6, 7, 8 are performed.

Adjustment of the period and peak value of approximate functions: This section introduces how to give a reference movement for the impaired knee after the start time of a phase, mainly including the adjustment of T and A of the current phase based on the measured real-time data of the healthy limb.

In each sampling point, it is required to determine whether or not T will be adjusted before outputting a reference angle for the impaired knee. Since the approximate functions expressing the movement of the healthy knee has been described for the 8 phases (function expressions correspond to 'ipsilateral' in Table 2, the period ($T_{\_healthy}$) and peak value ($A_{\_healthy}$) of the function that expresses the movement of the healthy knee can be calculated by using two knee angles measured in two sampling points. However, if the first two angles of the current phase are applied for calculating $T_{\_healthy}$ and $A_{\_healthy}$, it is easy to get large errors. Therefore, $T_{\_healthy}$ and $A_{\_healthy}$ are calculated after two or more integral times of 'interval_point_Num' sampling points. Currently, 'interval_point_Num' is defined as one half number of the sampling points of the phase in the first half of the gait cycle. By applying this method, the calculated $T_{\_healthy}$ will be used as the new T of the approximate output function (expressing the movement of the impaired knee) directly. The calculated $A_{\_healthy}$, together with the start and finish angles recorded in the first 8th phase are used to regulate A of the approximate output function. However, before performing this adjustment, the cycle time, start angle and finish angle of the phase in the first half of the gait cycle will be used to calculate T and A of the approximate output function (the method introduced in 2.1 section).

Since the function expressions of the healthy knee in the phases of foot flat, mid-stance, earlier-swing and later-swing are the same (sine function, refer to Table 1), the same method is applied for calculating the $T_{\_healthy}$ and $A_{\_healthy}$. Simultaneously, as for the phases of earlier-stance, later-stance, pre-swing and mid-swing (the corresponding function expressions are cosine function), the same method is applied for calculating the $T_{\_healthy}$ and $A_{\_healthy}$ as well. The methods in these two cases are explained as following: In which $\Delta\alpha 1$ denotes the angle difference between that obtained in integral times of 'interval_point_ Num' sampling point and the start angle of the current phase, accordingly, the corresponding time difference is denoted as Δt; when the time difference between the current sampling point and the start point equals to $2\Delta t$, the corresponding angle difference is recorded as $\Delta\alpha 2$.

The Function expression of the healthy knee in the current phase is a sine function:

$$\Delta\alpha 1 = A_{\_healthy}\sin\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right) \tag{1}$$

$$\begin{aligned}\Delta\alpha 2 &= A_{\_healthy}\sin\left(\frac{\pi}{2T_{\_healthy}}2\Delta t\right)\\ &= 2A_{\_healthy}\sin\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right)\cos\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right)\end{aligned} \tag{2}$$

By combining Eq. 1 and 2, $T_{\_healthy}$ and $A_{\_healthy}$ can be expressed as Eq. 3 and 4:

$$T_{\_healthy} = \frac{\Delta t\pi}{2\mathrm{acos}\left(\Delta\alpha 2 / 2\Delta\alpha 1\right)} \tag{3}$$

$$A_{\_healthy} = \frac{\Delta\alpha 1}{\sin\left(\pi\Delta t / 2T_{\_healthy}\right)} \tag{4}$$

where, acos denotes inverse cosine function.

The function expression of the healthy knee in the current phase is a cosine function:

$$\begin{aligned}\Delta\alpha 1 &= -A_{\_healthy}\cos\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right) + A_{\_healthy}\\ &= A_{\_healthy}\left(1-\cos\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right)\right)\end{aligned} \tag{5}$$

$$\begin{aligned}\Delta\alpha 2 &= -A_{\_healthy}\cos\left(\frac{\pi}{2T_{\_healthy}}2\Delta t\right) + A_{\_healthy}\\ &= 2A_{\_healthy}\left(1+\cos\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right)\right)\left(1-\cos\left(\frac{\pi}{2T_{\_healthy}}\Delta t\right)\right)\end{aligned} \tag{6}$$

By combining Eq. 5 and 6, $T_{\_healthy}$ and $A_{\_healthy}$ can be expressed as Eq. 7 and 8:

$$T_{\_healthy} = \frac{\Delta t\pi}{2\mathrm{acos}\left(\Delta\alpha 2 / 2\Delta\alpha 1 - 1\right)} \tag{7}$$

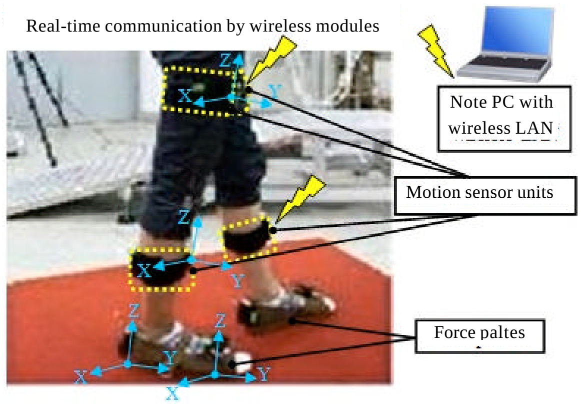

Fig. 5: 3D motion sensor units and instrumented shoes were used for detecting gait information and foot reaction force, respectively

$$A_{\_healthy} = \frac{\Delta\alpha 1}{(1-\cos(\pi\Delta t/2T_{\_healthy}))} \quad (8)$$

No matter in which case, $T_{\_healthy}$ will be considered as the new period of the approximate output function and the new A of the approximate output function is adjusted as:

$$A = A \pm A\_healthy\text{-}(\alpha_{\_finish}\text{-}\alpha_{\_start}) \quad (9)$$

where, $\alpha_{\_finish}$ and $\alpha_{\_start}$ represent the finish angle and start angle of the healthy knee recorded in the first 8th phase. If the motion trajectories of the two knees have the same motion direction in one phase, plus sign will be used instead of '±', whereas if the motion trajectories of the two knees have a reverse motion direction in one phase, minus sign will be used instead of '±'.

Experiment design: Walking experiment was carried out on four subjects (three males and one female, the average age is 26.5 years old). Each subject performed walking task 5 seconds (about three or four steps). During the process of walking, 3D motion sensor units (INSENCO Co., Ltd, Jiaxing, China) were used to measure orientations of the shank and the thigh of two legs and instrumented shoes (INSENCO Co., Ltd, Jiaxing, China) were worn by the subjects to measure triaxial Ground Reaction Force (GRF, including forefoot reaction force and heel reaction force) and foot segments' orientations (Fig. 5). The four motion sensor units and a pair of instrumented shoes transferred sensor data to a Personal Computer (PC) by a wireless local area network with the sampling frequency of 100Hz. All the sensors' raw data can be saved in a text file. In order to achieve the angle and angular velocity of the knees, knee joints' off-line kinematic analysis was implemented based on the measurements of 3D segment orientations of the shank and the thigh, refer to the method introduced by Liu *et al*., 2010, 2011).

To simulate real-time detection of GRF and knee joints' kinematic information (angle and angular velocity), the forefoot reaction force, heel reaction force, knee angle and angular velocity of one limb (represents the healthy limb) were stored in four buffers, respectively. Then, during the process of software verification, one set of data (values corresponding to the above four parameters) in each sampling point were considered as current input information. By using this input information and the proposed method, a reference angle of the contra-lateral knee (represents the impaired knee) is given in real time.

RESULTS AND DISCUSSION

As for the four subjects, the accuracy of gait phase identification was 100% and the calculated reference movement matched the movement of the other knee. One representative result is given Fig. 6, in which the blue line represents the measured angle information of the right knee and the green line represents the reference movement of the left knee given by using our proposed method. Simultaneously, the practical motion trajectory (blue line) as well as the calculated motion trajectory (green line) of the left knee is given in Fig. 7. It can be seen that the reference movement given by using our proposed method matches the knee movement of the other limb. Even though the calculated reference movement and the practical movement of the same knee has a clear difference, it is acceptable because the two

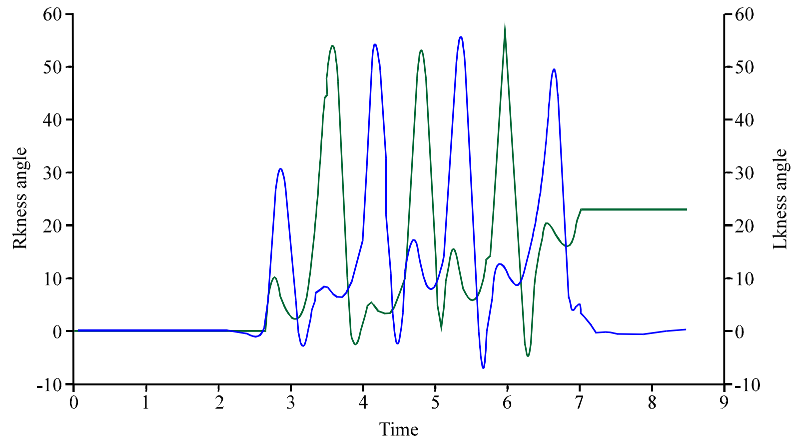

Fig. 6: Detected angle of the right knee (blue) and the output angle of the left knee (green)

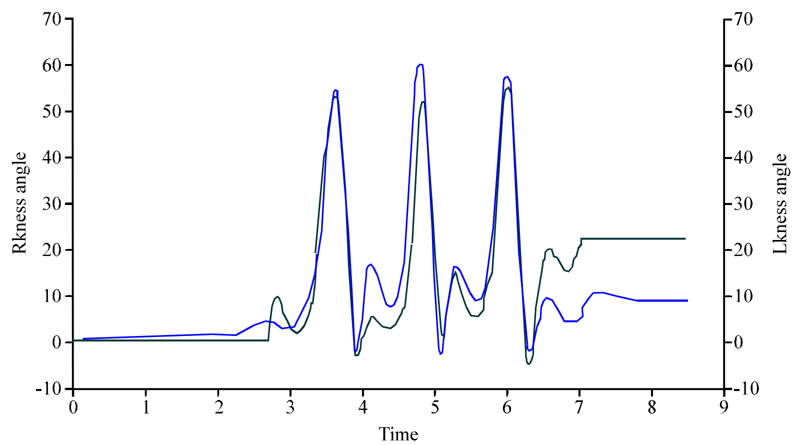

Fig. 7: Comparison between the detected angle (blue: Lknee Angle1) and the calculated reference angle (green: Lknee Angle2) of the left knee

limbs do not achieve fully symmetrical movement in the actual process of walking, that is, the motion trajectories of the two limbs with an interval of half a gait cycle also has a clear difference. Overall, the results verify the feasibility of our proposed method.

This study proposed a method to calculate reference movement of the impaired knee in real-time, which will be favorable to provide a correct reference movement for an artificial limb or exoskeleton prosthesis in the impaired limb side. According to the movement characteristics of periodic repeatability and symmetry of bilateral lower limbs during the process of walking, one gait cycle is divided into 8 phases: Foot flat, earlier-stance, mid-stance, later-stance, pre-swing, earlier-swing, mid-swing and later-swing, then, the phases of foot flat, earlier-stance, mid-stance and later-stance of one limb correspond to the phases of pre-swing, earlier-swing, mid-swing and later-swing of the other limb, respectively. Thus, the reference movement of a person's impaired knee can be calculated based on the movement of his own healthy knee. This is favorable to enhance movement symmetry of the two limbs. And if the calculated reference movement is used for controlling an artificial limb or exoskeleton prosthesis in the impaired limb side, the artificial limb or exoskeleton prosthesis could assist patients to achieve smooth movement more easily.

In addition, different phases are identified in view of the characteristics of peak value, valley value, zero value, increasing or decreasing trend of the healthy limb's information rather than some numerical values. Therefore, some change in stride length or walking pace does not affect the accurate identification of a new gait phase. That is, this method is favorable to enhance the accuracy of phase identification and further to present a continuous reference movement for an artificial limb or exoskeleton prosthesis in the impaired limb side.

Furthermore, sine or cosine functions are applied for expressing the knee movement in each phase approximately and the period and peak value of the approximate output function in the current phase is recalculated based on the measured angles of the healthy knee. Thus, the reference movement of the impaired knee could adapt to changes in stride length and walking pace. This is preferable to control an artificial limb or exoskeleton prosthesis in the impaired limb side to work with higher intelligence and stability and to assist patients to walk smoothly.

However, in the current verification experiment, the number of subjects was small. In order to obtain statistically significant results, further experiment should be performed on many more subjects. Otherwise, cable 3D motion sensor units and instrumented shoes should be applied for achieving a set of data in each sampling time rather than a text file after an interval of time. And an embedded system should be designed for further verifying the proposed method.

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