An IMU-Based Gait Detection for a Wearable Walking Assistive Device

Shanhai Jin, Changfu Jin, Xiaodan Wang, and Yonggao Jin

***Abstract*—In implementation of walking assistive devices, one major challenge is to detect gait cycle exactly, and then provide assistive force at proper timing in the gait cycle. This paper presents an IMU-based gait detecting algorithm for a wearable walking assistive device. The presented algorithm detects gait cycle by using measured hip velocity and hip angle signals. In addition, it is computationally inexpensive, and thus suitable for real-time applications. Experimental results on two subjects validated the effectiveness of the presented algorithm.**

***Index Terms*—IMU, gait detection, real time, wearable assistive device**

I. Introduction

It is reported that the quality of elderly life is positively linked to physical activities [1], [2]. However, due to the age-related muscle decline, many elderly persons have fewer and shorter physical activities compared to young ones [3]. On the other hand, decrease of physical activities, in turn, induces further muscle degeneration.

Recently, in order to prevent abovementioned vicious cycle, numerous exoskeletons for elderly persons have been developed. For example, a walking rehabilitation device for lower limbs presented by Cyberdyne Inc. [4]–[6]. As another example, a walking assistive device for improving walking function presented by Honda Motor Co., Ltd. [7]–[9]. The major advantage of the exoskeletons is that it can support the entire or a portion of body weight through rigid frames. In addition, they can generate a sufficiently large assistive force for the lower limb joints. However, in the case of exoskeletons, misalignment of axial joints between exoskeleton and wearer may produce an uncomfortable force [10]. In addition, the movement range of lower limbs is constrained by rigid frames.

To avoid drawbacks of exoskeletons, the author's group is engaged in developing a soft wearable walking assistive device for elderly persons [15]–[17]. The device provides small but effective assistance force for hip flexion. In addition, it is lightweight, and it does not constrain the movement range of lower limbs.

For implementing walking assistive device including ours, one major challenge is to detect gait cycle exactly, and then provide assistive force at proper timing in the gait cycle. A various of gait cycle detection methods have been utilized. For example, the author's group [11] have adopted bending sensors to measure hip angle for detecting gait cycle. However, the reliability of the measured hip angle depends on attaching position of the bending sensor. Moreover, the cost of a bending sensor is high, and the accuracy reduces with aging. As an alternative way, some researchers [12]–[14] have used foot switches. However, a foot switch requires long cabling between the attached position of shoes and the controller. Furthermore, the accuracy of a foot switch also degrades with aging.

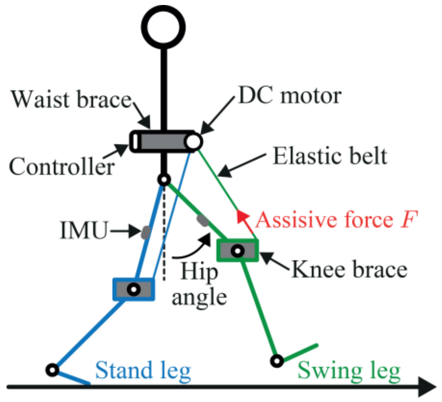

Fig. 1. Overall structure of a soft wearable walking assistive device.

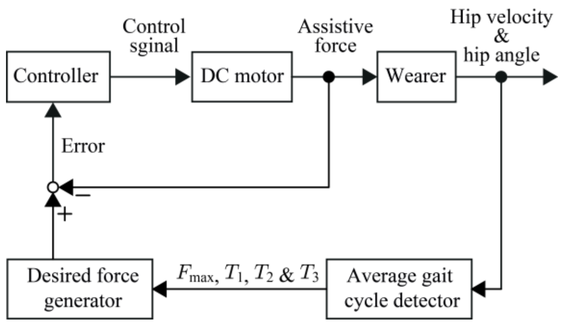

Fig. 2. Block diagram of the force control scheme.

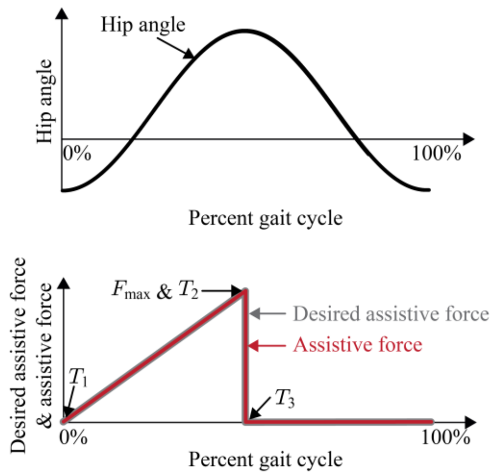

Fig. 3. Illustration of hip movement, desired assistive force and assistive force. Here, F_{max}, T_1, T_2 and T_3 represent maximum assistive force, start point, maximum assistive force point and end point, respectively.

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This paper presents an IMU-based gait detecting algorithm for the walking assistive device [15]–[17]. The presented algorithm detects gait cycle effectively by using measured hip velocity and hip angle signals. In addition, it is computationally inexpensive, and thus suitable for real-time applications.

The rest of this paper is organized as follows. Section II describes a soft wearable walking assistive device. Section III presents an IMU-based gait detecting algorithm, and Section IV shows experimental results for validating the effectiveness of the presented algorithm. Finally, Section V concludes the paper.

II. Soft Wearable Walking Assistive Device

In Refs. [15]–[17], the author's group presented a wearable walking assistive device for elderly persons. Fig 1 shows the overall structure of the assistive device. For each leg, the device consists of one DC motor and one control unit that attached to a waist brace, one knee brace, one elastic winding belt and one IMU sensor. In swing phase, the motor winds up the belt, and the resultant force generated on the belt is transmitted to the wearer for assisting the hip flexion. The major advantage of the device is that, excluding the DC motor and the control unit, it is composed of soft material, and thus safe and compatible interaction can be realized with the wearer. In addition, the elastic winding belt can absorb undesirable reaction forces in cases of large control errors or disturbances.

Fig. 2 shows control scheme of the device. Hip motion information and generated assistive force are measured and transmitted to the control unit. In the controller, gait cycle is calculated according to the hip motion information. Average gait cycle is obtained by using the last three cycles, and it is converted to a percentage scale of 0% to 100%. Four parameters $F_{\max}$, T_1, T_2 and T_3, which represent maximum assistive force, start point, maximum assistive force point and end point, respectively, are used in generating desired assistive force trajectory, as shown in Fig. 3. A force feedback control scheme is employed for tracking the desired assistive force.

III. Gait Detecting Algorithm

This section presents an IMU-based gait detecting algorithm for the walking assistive device [15]–[17].

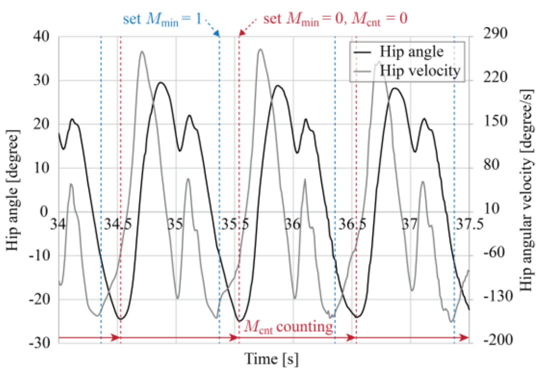

Fig. 4. Typical hip velocity and hip angle data.

Fig. 4 shows typical hip velocity and hip angle data. It is shown that both velocity and angle produce periodic movement. Thus, it is possible to obtain gait cycle from hip motion information.

Now, the task is to choose an appropriate point, which is easy to be detected, as the start point of gait cycle. It is known that, for a periodic signal, two extreme points of each period are always the first choice because of their simplicity to be detected. Thus, for a gait detecting algorithm, the events of maximum hip angle and minimum hip angle are the best two candidates for the start point of gait cycle. However, in the case of using maximum hip angle, due to the valley and hill shapes just after the maximum hip angle caused by heel contact, it is not easy to distinguish the two successive peaks. Moreover, because of the fact that the device provides assistive force for the hip flexion, it is desirable to set the start point as the start timing of swing leg, i.e., the event of minimum hip angle. Based on this rationale, the event of minimum hip angle is selected for segmenting hip cycle.

Now, the task becomes how to search the minimum hip angle. By careful observation on hip angle, one can see that, before the event of minimum hip angle, the values of hip velocity and hip angle are both negative. In addition, the value of hip angle decreases monotonously. Thus, it is preferable to implement the process of searching minimum hip angle just after the simultaneous detections of negative hip velocity and monotonic decrease of hip angle in negative value. Then, the event of minimum hip angle is found by searching valley point in hip angle, and the gait cycle between two consecutive events of minimum hip angle is recorded by the controller. Estimated gait cycle is obtained by averaging the last three recorded cycles. After that, estimated gait cycle is converted to a percentage scale of 0% to 100%.

As a whole, the complete algorithm of gait detecting is given as follows:

Algorithm IMU - based gait detection

if $\omega(k)<0$ **and** $\theta(k)<0$ **and** $M_{\text{cnt}}>N_1$ **then**

  if $M_{\min}==0$ **and** $(\omega(k-N_2)<\omega(k))$ **then**

    $M_{\min}=1$

  end if

end if

if $M_{\min}==1$ **and** $\theta(k-2N_3)>\theta(k-N_3)$ **and** $\theta(k\text{-}N_3)<\theta(k)$

  $c(i)=M_{\text{cnt}}-N_3$

  $c_{\text{ave}}=\sum_{j=0}^{2}c(i-j)/3$

  $c_{\text{coef}}=\dfrac{c_{\text{ave}}}{100}$

  $M_{\text{cnt}}=0,\ M_{\min}=0$

end if

$c_{\%\text{cur}}(k)=M_{\text{cnt}}/c_{\text{coef}}$

$M_{\text{cnt}}=M_{\text{cnt}}+1$

return $c_{\%\text{cur}}(k)$

where k and i denote discrete time index and gait cycle index, respectively. In addition, $\theta(k)$ is the hip angle, $\omega(k)$ the hip angular velocity, $c_{\%\text{cur}}(k)$ is the current gait cycle expressed in percentage scale, N_1, N_2 and N_3 are threshold values, and M_{cnt}, M_{min}, $c(i)$, c_{ave} and c_{coef} are intermediate variables.

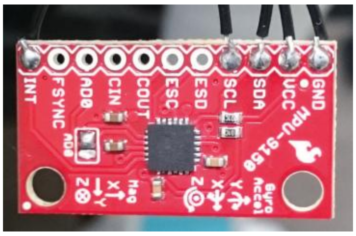

Fig. 5. IMU sensor: MPU-1950 (InvenSense). It consists of a 3-axis accelerometer, a 3-axis gyroscope and a 3-axis magnetometer.

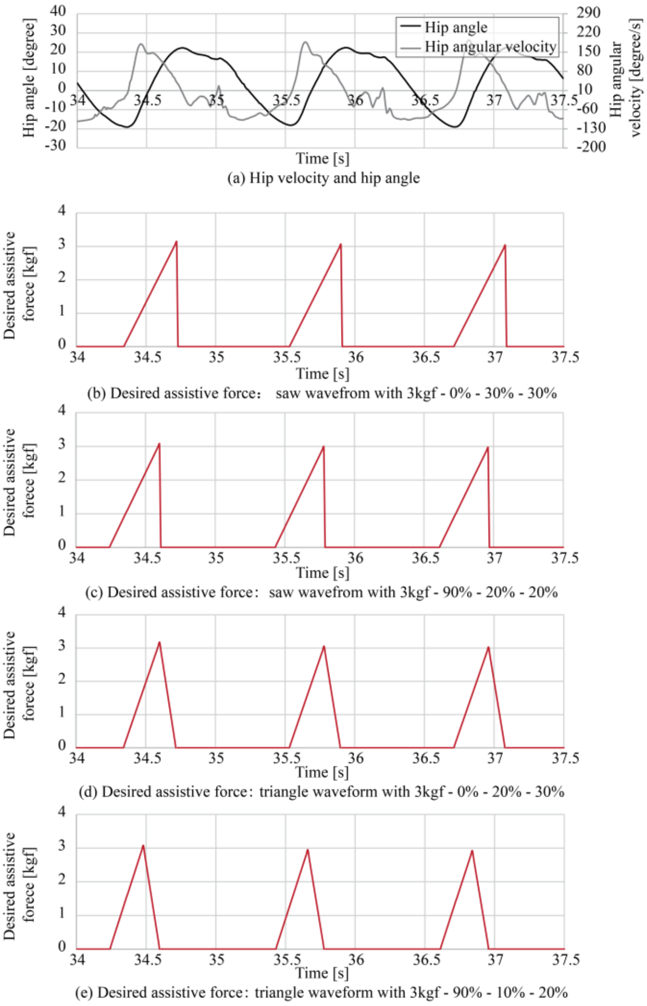

Fig. 6. Results of subject 1. (a): hip velocity and hip angle; (b) and (c): saw waveform desired assistive forces; (d) and (e): triangle waveform desired assistive forces.

IV. Experiment

This section experimentally validates the effectiveness of the presented algorithm. An IMU sensor (MPU-9150, InvenSense), as shown in Fig. 5, was used in the experiments. The sensor was equipped at the posterior of the thigh. Hip velocity in sagittal plane was measured by the sensor at sampling interval $T = 0.001$ s was in discrete time, and it was transmitted to a 16-bit microcontroller (dsPIC33F, Microchip Technology Inc.). Hip angle was calculated by integrating the measured velocity signal. The presented gait detecting algorithm was implemented by using the obtained hip velocity and hip angle signals.

Two healthy subjects participated in the experiments. For each subject, four desired assistive force trajectories, of which parameters were described in $F_{max} - T_1 - T_2 - T_3$, form, were generated by using the presented algorithm. Among them, two of them were saw waveforms with parameters 3kfg–0%–30%–30% and 3kfg–90%–10%–20%. The other two were triangle waveforms with parameters 3kfg–0%–30%–30% and 3kfg–90%–10%–20%. In addition, the three thresholds were set as $N_1 = 10$, $N_2 = 2$ and $N_3 = 3$, respectively.

Fig. 6 and Fig. 7 show experiment results. One can observe that the presented algorithm reliably detected minimum hip angle and estimated average gait cycle in percentage scale. It should be noticed that, the algorithm is sufficiently robust for different gait pattern, i.e., subject 1 and subject 2. The figures also show that, based on the results of the algorithm, four different desired assistive force trajectories were successfully generated for each subject. This can be attributed to the effectiveness of the presented algorithm.

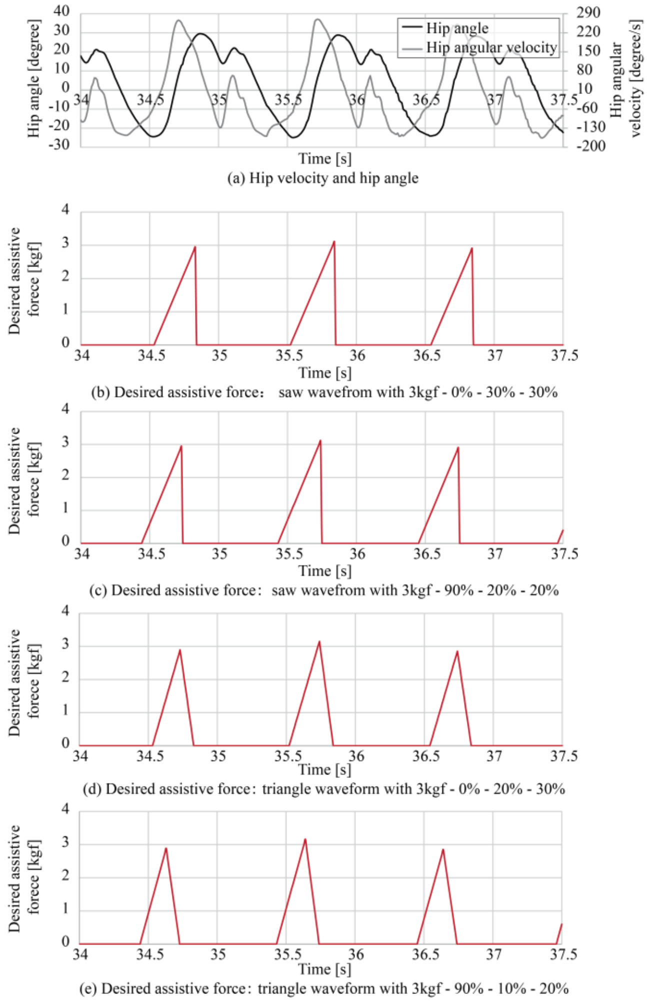

Fig. 7. Results of subject 2. (a): hip velocity and hip angle; (b) and (c): saw waveform desired assistive forces; (d) and (e): triangle waveform desired assistive forces.

V. Conclusions

This paper has presented an IMU-based gait detecting algorithm for a walking assistive device. The presented algorithm detects gait cycle by using measured hip velocity and hip angle signals. In addition, it is computationally inexpensive, and thus suitable for real-time applications. Experimental results on two subjects validated the effectiveness of the presented algorithm. As a future work, it is expected that the presented algorithm can be further extended for detecting gait pattern in real time.

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