AN AUTONOMOUS CONTROL SYSTEM FOR A PROSTHETIC FOOT ANKLE

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Abstract: When walking on inclined ground the biological foot adjusts the ankle angle accordingly. Prosthetic foot users have often a limited range of motion in their ankle which makes walking on hills uncomfortable. This paper describes a system which can autonomously correct the ankle angle to the inclination. The ground angle is estimated using an accelerometer. The angle foot blade to heel is then adjusted with a DC-motor. Since the controller only activates the motor when the foot is lifted and thus not loaded, a small powered system can be used.

Keywords: Accelerometers, hill climbing, walking, actuator, adaptive system

1. INTRODUCTION

Prosthetic feet design has advanced in the last two decades giving amputees, but otherwise healthy people, nearly full range of walking capabilities (Marks and Michel, 2001). Many new feet are based on carbon fibre springs which have the capability of storing and releasing energy during walk. This enables damping at heel down and push off at foot lift. Furthermore, systems with spring loaded chock absorber have been introduced. But the existing systems are still limited in their capability of adapting to new inclining circumstances i.e. hills or heel hight variation.

Studies by Leroux *et al.* (2002) and McIntosh *et al.* (2005) showed that when walking in an inclination, a tilting of the trunk was necessary to help in creating a forward momentum. This tilt increased with inclination angle almost exponentially. The largest flexion differences in the joints, compared to level walking, occur from mid-swing to early stance(heel down). But the ankle rotation between stance and swing (push-off) is not effected while the ankle difference is almost constant during heal down and weight acceptance. Thus, the lower limb angle to a vertical line is affected very little by the inclination (Leroux *et al.*, 1999).

In downhill, compared to level walking, the joints have to absorb more energy caused by the combined forward and downward movement. This can be seen by a shorter stride length (Leroux *et al.*, 2002) and is compensated, mostly by the knees and somewhat by the hip, in the end of the stance-phase. At the early stance phase when the foot is put down a larger momentum is used foremost in the knee but also in the ankle (Redfern and Dipasquale, 1997). In the ankle angle there is an increase in dorsiflexion starting just before heel down and a decrease of plantarflexion at late stance (Kuster *et al.*, 1995). At toe-off the change is in the same size as the hill inclination angle. Also in the swing phase the flexion is smaller.

McIntosh *et al.* (2005) stated that people with reduced range of motion i.e. elderly and prosthesis users, these changes in flexion are not possible. Many prosthetic feet have fixed ankle position and attempting to move the body's center of mass forward may cause *instability* when walking on inclinations.

Active control of ankles are done in orthotic devices as well as in artificial ankles. The orthoses are used in rehabilitation of drop foot control. Both orthotic control with an electrical motor (Blaya and Herr, 2004) and orthotic rehabilitation device using pneumatic actuators (Gordon *et al.*, 2005) have been proposed. While the former still has to be designed with a realistic transducer the latter is limited to hospitals or labs with compressed air facilities.

Gait, both for analysis and active control, is typically divided into 2-5 different phases: *Stance*, (*Heel down*, *Foot flat*, *Heel rise*, *Toe-off*), *Swing*. Automatic classification of gait phases has previously been done using various wearable sensors e.g foot resistive sensors (FSR), gyros, accelerometers or goniometers (Ng and Chizeck, 1997). The objective has mainly been to electrical control in assisting people with drop foot. FSRs can, although being non linear, be used as foot switches inside shoe soles e.g. (Smith *et al.*, 2002; Skelly and Chizek, 2001). But tests on parapledic subjects showed that they were limited when classifying foot flat. Instead Pappas *et al.* (2001) combined a gyroscope with 3 FSRs. The gyro signal was integrated to get the foot angle. This angle and the angular velocity together with the information from one of the FSR sensor give the actual state. This was shown to work on varying ground conditions. Aminian *et al.* (2002) only used a gyro positioned on the shank. They utilized that the shank angular velocity have distinct signal features at heel-strike and toe-off. They also found that the gyro has several advantages versus accelerometer: 1) The gyro is not sensitive to placements (Tong and Granat, 1999) as long as it is parallel to the mediolateral axis. 2) The angular velocity is less noisier since the accelerometer is a differentiated velocity. 3) The rotation can easily be estimated by integration and provides thereby spatial information. The major drawbacks though, are the sensitivity for shock due to the mechanical fastening of the beam inside the gyro. It also requires signal processing removing the internal drift. Thus, it can not be used to measure ground angle.

Williamson and Andrews (2000) mounted three accelerometers on the shank. With machine learning algorithms different 5 states could be detected. But Willemsen *et al.* (1990) found that level detection could not be used in raw accelerometer data. Instead they cross correlated the signals and differentiations with normalized course using an estimated mean and standard deviation.

Studies have shown that moving the center of gravity using hip and knee, robots can be stabilized during gait and standing in inclinations (Takuma *et al.*, 2005; Zheng and Shen, 1990). Active ankles have been used to control torque with pneumatic actuators (van der Linde, 1999; Guihard and Gorce, 1996) or DC-motors (Yi and Zheng, 1996). The feet have also been extended with controllable toe joints (Nishiwaki *et al.*, 2002) in reducing knee joint load during high step climbing. Although much research has been done in the area of artificial muscles, results have not transferred easily to robotic design. Many artificial muscles suffer from being either small but weak or strong but bulky. Klute *et al.* (2002) proposed a combination of existing actuators giving a system which could more closely model the muscle-tendon dynamic behavior in both force-length and force-velocity. Especially that in the Achilles tendon. A recent trend is to investigate if electroactive polymers potential to be used as artificial muscles see e.g. (Bar-Cohen, 2001).

A major practical problem when trying to model an ankle actively is the power needed. We propose an approach where the dynamics is mostly handled by mechanical solutions and only adjusted by an active device. In this paper, a new actuator for foot angle adaption is presented. This is controlled autonomously by one single accelerometer. This choice of sensor enables both gait phase detection and ground slope measurement.

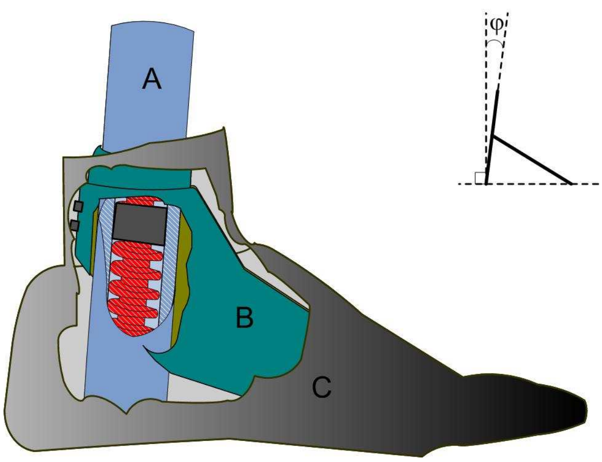

Fig. 1. By moving the footpad B along the heel axis A the ankle angle φ is changed. The mechanics is encapsulated in a protecting sock C.

2. NEW ACTUATOR DESIGN

The studied mechanical prosthetic design has ankle angle which is altered by moving the foot blade up or down (U.S. Patent 5.571.210) as shown in Fig. 1. The feature of this design is that length knee to heel is kept constant.

2.1 Mechanical design

A controller updates the mechanical behavior of the prosthesis when it is not loaded. Thereby, a weaker and less power consuming actuator can be used. Inside the heel axis is a DC-motor which moves the footpad up and down. The rotational-linear movement conversion comes from a ball screw. This is chosen to reduce the load on the motor to act only at adjustment phase. The position and thereby ankle angle is measured with a rotational encoder.

2.2 Control unit

All electronics are embedded in the prosthetic device. The digital controller is implemented in a low cost, off the shelf, 40 MHz PIC18F microprocessor. Signals are sampled at 50 Hz with 10 bit AD converters. One two directional ADXL311 accelerometer from Analog Devices is mounted on the inside of the protecting sock at the corresponding fifth metasarsal bone (In front of the C in Fig. 1). Only one direction of the accelrometer directions is used in this approach. The signal is analog filtered via 50 Hz first order low-pass filter. A current controlled DC-motor is feed from a 10 bit DA-converter. Furthermore, to analyze the system performance online, a bluetooth unit enables logging of data to a PC.

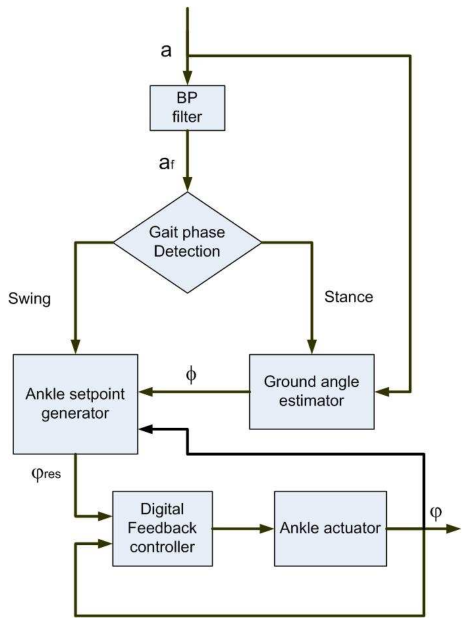

Fig. 2. Ankle control system.

3. ANKLE CONTROL SYSTEM

The feature of the proposed ankle control system is that it measures the ground angle of inclination and adjusts the ankle angle during the non loaded phases when the DC motor only is loaded with the footblade weight. The strategy is to use the swing phase for controlling and the stance phase for measuring as shown in Fig.2.

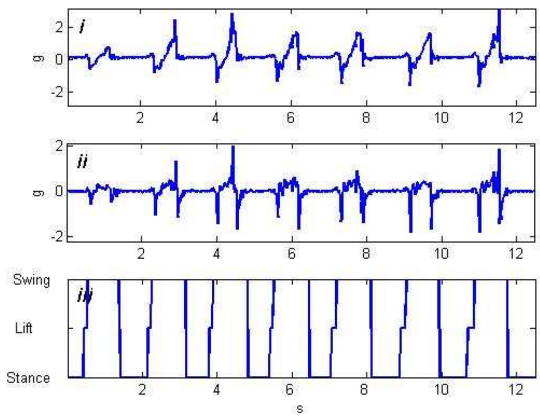

Fig. 3. Typical gait: *i*) The accelerometer signal. *ii*) Band pass filtered accelerometer signal. *iii*) Detected gait phases.

3.1 Gait phase detection

The gait is divided into three phases: *Swing Stance*, and *Foot lift*. That is; a control phase, a measuring phase and a phase where neither can be done. The separation between these phases is done at small foot angels. Thus most information is found in the forward-backward acceleration. Typical variations during normal prosthetic gait is shown in Fig. 3. This acceleration a, is band pass filtered removing high frequency noise and stationary angel dependency as:

$$a_f(k) = h_1 a_f(k-1) + a(k) - a(k-1) \quad (1)$$

This signal is used to switch between the three phases (also shown in Fig. 4):

Foot Down is when the acceleration is constant during several sequential samples i.e. a_f only consists of noise. This is realized by comparing a window w, of the signal with a threshold γ_s.

Heel Lift ends constant acceleration phase by a upward rotation of the foot blade.

Toe off ends this rotation and brings the foot into a forward swing and is detected by comparing the signal with a threshold γ_{To}.

3.2 Inclination measurement

In the stationary case when the foot is at inclination the angle ϕ is estimated from the acclerometer signal.

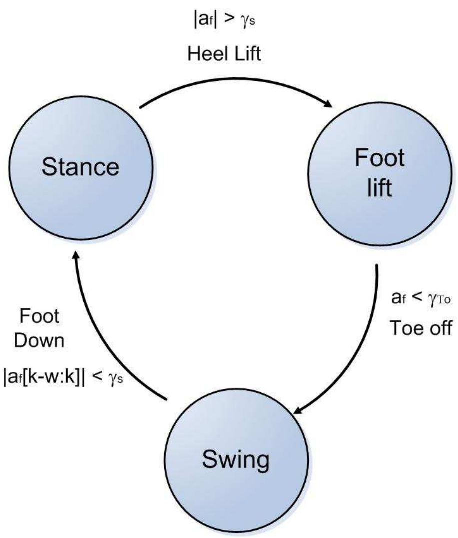

Fig. 4. Gait phases. Switching between these is by thresholding the band pass filtered accelerometer signal

$$a = g \sin(\phi) \quad (2)$$

where g is the gravitational constant.

3.3 Control design

The ankle controller consists of two levels: A high level ankle angle adapter and a low level ankle actuator.

The control law regulates the ankle angle with the inclination angle as setpoint. This was realized by a stepwise updating and thereby suppressing the sensitivity for e.g. small hills or irregularities in the surface. The incremental step of φ_{ref} is:

$$\Delta\varphi_{ref} = \begin{cases} d \cdot sign(\varphi - \phi), & |\varphi - \phi| > d \\ 0 & else \end{cases} \quad (3)$$

where d is a chosen threshold. The DC-motor of the ankle actuator includes a digital controller with the encoder signal feedback. This enables servo control and possible load variations caused by friction. But when the footblade hits the ground, either by a shorter stop than anticipated or irregularities of the surface, the actuation has to stop. This is handled by an included motion stop detector which compares the position φ(k) with the controller output u(k) as:

$$\left.\begin{array}{c} u(k) = u(k-1) \neq 0 \\ \varphi(k) = \varphi(k-1) \end{array}\right\} \Rightarrow Stop \quad (4)$$

4. MEASUREMENT SETUP

In the design stage, for safety reason, it is preferable to evaluate new principles without doing tests on people with amputees. Therefore, a prosthesis

Fig. 5. Prosthesis mounted on an evaluation socket.

evaluation socket was used(see Fig. 5). This is similar to the device previously studied by (Takahashi *et al.*, 2004) for natural walking analysis in different environments. The evaluation was performed by letting a healthy man walk up and down a hill and in a corridor at a self selected speed.

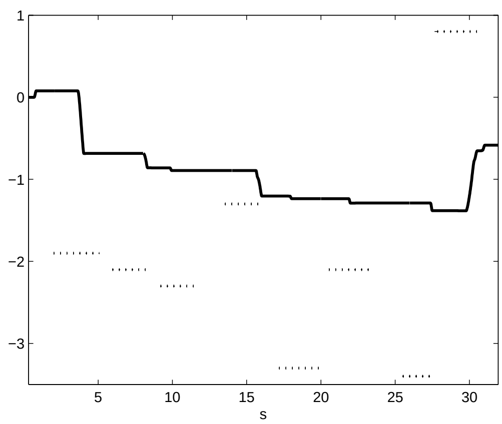

Fig. 6. Control performance walking on a hill. Estimated ground angle φ(dashed) from estimation phase. Resulting ankle angle ϕ(solid)

5. RESULTS AND DISCUSSION

Our preliminary test showed that the three necessary states were detectable by the use of one single bandpass filtered accelerometers signal. The test were done on typical gait conditions and showed that there is a characteristic toe off in the signal. Although more studies should be done to see how speed, ground conditions and non forward walking foot movement influence the classification approach.

In Fig.6 are the angles of consecutive eight steps(Seven downwards the hill and one ground level) shown. The stance phase is the time between the dotted lines which results in the dotted estimation angle. The slope was approximately 2 degrees but both irregularities and not having foot flat exactly in the direction of inclination caused variations seen in step 4,5 and 7. Several studies (Rehbinder and Hu, 2004; Sabatini *et al.*, 2005; Svensson and Holmberg, 2005) have previously shown that a system of gyros and accelerometers placed on the foot can be used to estimate with 1-2 degree accuracy. Our indoor experiments show similar accuracy of 0.5-1 degree. It was found that this accuracy was restricted by the inwards/outward inclining of the shank and that the prosthesis not was fixated to the protecting sock causing motion inside the shoe.

The accuracy of estimation was used as threshold level d in eq.(3). The performance of the adaptation is depending of the accuracy of the inclination estimation. But the actuator control design softly adjusted the ankle angle towards the ground angle. This setpoint was not always reached since the swing time varied. In the first and last step the setpoint was reached while in those in between the force stop eq.(4) became active. By using threshold the system becomes slower but also less sensitive to short variations. After five steps the ankle angle φ reached the ground angle $\phi \pm d$. Variations in load force caused by the surrounding sock and shoe saturated the control current of the DC-motor which should be adjusted in a future design.

6. CONCLUSION

A new system for autonomous ankle angle control has been developed. Ground inclination is estimated and used as setpoint for the angle controller. Adjustments are being made only when the foot is unloaded thereby requiring less energy. A hierarchical control system is designed with an inner lop controlling the ankle angle and a outer loop for ground slope tracking. The outer loop is based on only one single accelerometer sensor. This signal is used both for categorization of the swing and stance phase, and for estimation of ground slope.

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