STANCE-CONTROL OF AN ROBOTIC KNEE ORTHOSIS BASED ON ADMITTANCE CONTROLLER FOR GAIT ASSISTANCE

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Abstract— Stance control to assist the knee joint of people with mobility limitations can be implemented at a robotic orthosis using impedance/admittance controllers, which requires methods to obtain a smooth block-unblock of the knee joint during the gait cycle. This work proposes an admittance controller for a knee active orthosis, which adjusts impedance and damping parameters based on gait velocity and gait phases. The system was validated in volunteers and the results show good recognition of the gait phases and smooth adjustment of knee impedance. The modulation of the admittance parameters can be adjusted, and this orthosis can be adapted for cases with gait asymmetries.

Keywords— Admitance control, footswitch, gait cycle

Resumo— O controle de estabilidade da articulação do joelho de pessoas com problemas de mobilidade pode ser implementado utilizando controladores de impedância/admitância, o que requer métodos para obter um bloqueio-desbloqueio suave da articulação durante o ciclo da marcha. Este trabalho propõe um controlador de admitância para uma ortesis ativa de joelho, a qual ajusta os parametros de inércia e amortecimento, com base na velocidade e nas fases da marcha. O sistema foi validado em voluntários, e os resultados foram aceitáveis para reconhecer as fases da marcha, com ajuste suave da impedância no joelho. Também foi verificada a modulação dos parâmetros de admitância, com a órtese podendo ser adaptada em casos de assimetrias da marcha.

Palavras-chave— Controle de admitância, footswitch, ciclo da marcha.

1 Introduction

Currently, the number of people that requires devices to assist movements of their lower limbs has increased considerably. Walking is more difficult in elderly population and also for persons that suffer gait impairments, in neurological disorders as stroke (Mahlknecht et al., 2013), (Balaban and Tok, 2014). These conditions often lead to injury, disability, risk of falls, loss of independence and reduction in the quality of life. In order to apply functional compensations during gait, knee orthoses with stance control (SC) are usually prescribed, which is a strategy designed to release the knee during the swing phase to allow free knee motion and keep the knee locked in full extension during stance phase (Ir and Azuan, 2015). It is reported that the SC strategy may be used to increase walking speed, improve gait kinematics (knee range of motion, stride and step lengths), reduce energy expenditure and gait asymmetry, to both affected and unaffected legs, and allow less stressed paretic musculature in patients with muscular weakness (Zissimopoulos et al., 2007), (Zacharias and Kannenberg, 2012), (Rafiaei M, 2016).

Some mechanisms and designs have been used to develop stance control using robotic orthosis with promise results (Ir and Azuan, 2015). However, some methods are required to improve the real-time gait phase recognition providing smooth switching operation between the stance and swing phases. Furthermore, control strategies to improve the human–robotic cooperation are also required. Human–robotic systems are intended to provide an effective human support through assisting the limited motor capability of the user (Tsuji and Tanaka, 2005). During walking, to allow a more natural gait, these aspects remain remarkable challenges to warranty a suitable response of a robotic orthosis using the SC principle (Yakimovich et al., 2009), (Ir and Azuan, 2015).

As the user changes his/her joint impedances by regulating the postures and the muscle-contraction levels to maintain the system stability during the movement, some methods have been proposed to design and control a human–robotic system through an impedance/admittance-controller. This may be used to implement a SC to provide adequate knee stability and allow a more normal gait. Also, this controller may offer the possibility of investigating knee impedance variations in human control performance, such as done by other studies focused on upper limbs (Tsuji and Tanaka, 2005).

An impedance controller allows regulating the mechanical impedance of a system, through the relation between force, position and its time-derivate. This relation is given by three components involved in the impedance: stiffness, damping and inertia. An admittance controller is one of the variations of the impedance controller and their performance is determined by precision of force sensor, actuator position precision and bandwidth. Admittance controllers are stable in high stiffness conditions, therefore more suitable for implementation of a SC, due to the high and stable stiffness needed to avoid knee collapse during stance phase (Espinosa, 2013).

The objective of this work is to propose a SC using an admittance controller and a method to online adjust of the impedance parameters to switch the knee impedance throughout the gait cycle. For this, an instrumented insole is used to identify the gait phases, and gait velocity, gait phases and user height data are used to calculate suitable times to increase or decrease inertia and damping parameters.

2 Materials and Methods

2.1 Instrumented insole

Figure 1 (a) shows the instrumented insole to detect gait phases, which was built in this work. Four force sensors were placed on the plantar surface of the foot, which produce voltages relative to the amount of force on the sensor. The sensors employed were FlexiForce A401, that are piezoresistive force sensors (FSR) with a sensing area of 25.4 mm and a standard force range of 111 N (0 - 25 lb).

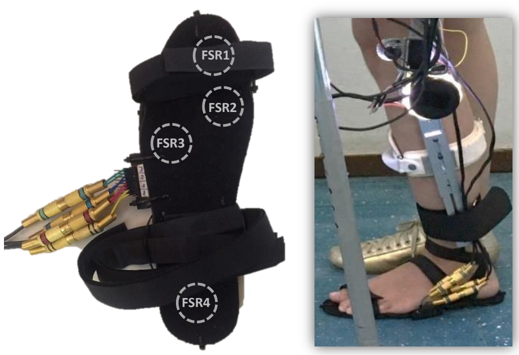

Figure 1: Instrumented insole and FSR locations.

Figure 1 shows the sensor locations which are based on the peak plantar pressure data reported in (Linah Wafai and Begg, 2015) and correspond to calcaneus, 1st and 5th metatarsals, and hallux bone. The signals were acquired through the Diamond-MM-32DX-AT Analog I/O Module of a PC-104 computer, and sampled at a frequency of 1 kHz.

2.2 Active knee orthosis

Figure 2 shows the knee active knee orthosis called ALLOR "Advance Lower Limb Orthosis for Rehabilitation", developed at Federal University of Espirito Santo (UFES/Brazil), which was employed to test the impedance controller.

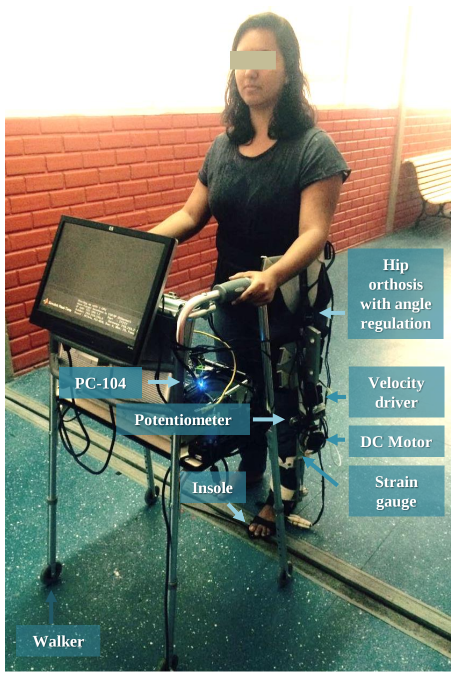

Figure 2: Setup to gait rehabilitation with Advance Lower Limb Orthosis for Rehabilitation (ALLOR).

This exoskeleton is mounted on the left leg of the user and is adaptable to different anthropometric setups, that include heights of 1.5 to 1.85 m and weights from 50 to 95 kg. The weight of ALLOR is 3.1 Kg and it provides both mechanical power to the knee joint and feedback information related to knee angle, interaction torque and gait phases. ALLOR was developed for knee rehabilitation in both sit position and during gait (in this case, the user must use canes, crutches or walker to obtain support). ALLOR includes a hip orthosis with manual angle regulation from 0 to 80 degrees of flexion and extension. Although this joint is not active, the regulation according to the user requirement allows to establish a safe range of motion. For users that required and active assistance in the hip, a functional electrical stimulation can be applied.

2.3 Gait phase detector

In order to recognize the gait phases with the information from the instrumented insole, a gait phase detector (GPD) in Matlab Simulink® real time was implemented. The signals were conditioned through a low-pass filter Butterworth of 5th-order, with cutoff frequency of 10 Hz. After, the signals were compared to a threshold of 0.5 V in order to obtain contact information (on-off) in a footswitch mode. Figure 3 (a) shows the combination of the on-off information from the sensors, which was used to generates a gait pattern, as shown in Figure 3 (b).

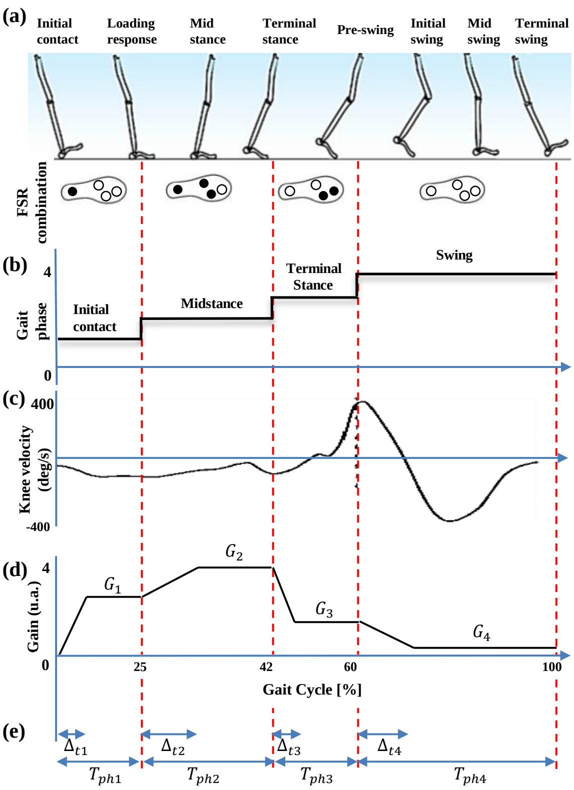

Figure 3: Events related to gait phases. (a) on-off sequence of FSR sensors throughout the gait cycle; (b) gait-phase pattern generated by the instrumented insole; (c) knee velocity throughout the gait cycle; (d) gain levels that are required to obtain a suitable impedance for knee support during gait; (e) time to decrease/increase gain values respect time of gait phases.

This pattern allows recognizing the sequence of the following phases: initial contact (IC) defined by the heel contact; mid-stance (MS) defined by a flat foot contact; terminal stance (TS) defined by the heel off; and swing phase (SW), defined by the foot off.

2.4 Admittance controller and parameters adjustment

The transfer function of an admittance-controlled Y(s) is given as:

$$Y(s) = 1/(Ms^2 + Ds + K), \quad (1)$$

where M, D and K represent the inertia, damping and stiffness of the end-effector, respectively. Then, we assume the use of a velocity controller in the robotic orthosis, therefore the admittance controller was performed through a transfer function of first order that relates the input force with other variables as torque, inertia, and damping, expressed as:

$$\dot{q}_a(s) = \tau(Ms + D)^{-1}, \quad (2)$$

where $\dot{q}_a$ is the output velocity, and τ is the interaction torque.

In order to adapt the joint impedance during gait cycle, the admittance parameters need to change with a smoother response. The objective is to allow a suitable knee velocity ((Figure 3 (c)), locking the knee joint only in the stance phase to resist knee flexion while allowing free knee extension and free knee motion in the swing phase, in order to allow free joint rotation in flexion and extension (Yan et al., 2015). For that, a gain G to increase or decrease the M and D values according to the current phase, recognized by the GDP, is required. Figure 3 (d) shows the levels of G throughout the gait cycle to obtain such variation, as follows:

$$M_i = M_d G_i, \quad (3)$$

$$D_i = D_d G_i, \quad (4)$$

where i is the phase number assigned (1 for IC, 2 for MS, 3 for TS and 4 for SW), M_d and D_d are the inertia and damping default values, respectively, with a ratio $M_d/D_d = 0.2$. The increase/decrease of G is executed in a time Δ_t, which depends on the period of each phase Tph. As shown in Figure 3 (e), a suitable Δ_t does not have to exceed the Tph. This value can be expressed as:

$$Tph_i = TGC(Ph_i)f_s, \quad (5)$$

where, TCG is the time of the gait cycle in seconds, Phi is the percentage of each phase respect the gait cycle, and f_s is the sampling frequency in samples per second. According to gait studies (Arnos, 2007), TGC can be estimated through Equation 6.

$$TGC = SL/v, \quad (6)$$

where, SL is the stride length in meters and v is the user velocity in meters per second. SL can be estimated with the value of user height in meters multiplied by the constant 0.826 (Arnos, 2007). Then, Tph, for each phase, can be expressed as:

$$Tph_i = (0.826/100v)HPh_i f_s. \tag{7}$$

Experimental tests to validate Ph_i with the instrumented insole were conducted. The following percentages for each phase: IC 16±4%, MS 38±6%, TS 6±0.8% and SW 40±4%.

IC and MS represent the more critical phases when a knee support is required, Δ_t should allows a time of stabilization of G value. Therefore, for this method, the Ph_i value was defined as: Ph_1=10%; Ph_2=20%; Ph_3=30% and Ph_4=40%. Ph_3 was choose in order to simplify the method, due to *TS* has short duration respect other phases, and does not allow stabilization of its G value. Considering that Δ_t represents 50% of its corresponding Tph, Equation 8 results:

$$\Delta_{ti} = (0.0413i/v)Hf_s. \tag{8}$$

Figure 4 shows the admittance controller implemented, which is based on Equation 2. This includes an outer force control loop implemented over a inner velocity control loop. Motor controller performs the velocity closed-loop control with information feed from Hall sensors on the motor structure. *Ph(t)* is the phase recognized by GDP.

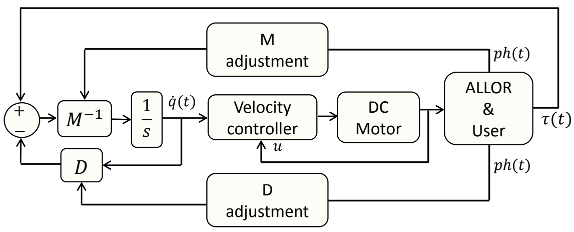

Figure 4: Admittance controller implemented.

2.5 *Experimental protocol*

A test to evaluate the G adjustment reflected in the variation of M parameter during gait was carried out. Seven healthy subjects (35±5.54 years; height 1.70±0.06 m) without lower limb injury or locomotion deficits participated of the test. The subjects walked a distance of 10 m at normal speed using the instrumented insole without the exoskeleton. Three trials were performed with the acquisition hardware attached to a four wheel walker, in order to the user have a mobile platform during the test. For our approach, the gain G for each sub-phase was 1 for IC, 3 for MS, 5 for TS and 10 for SW. Then, to evaluate the impedance control, two healthy subjects (1 male 35 years; height 1.70 m; weight 84.5 kg, and 1 female 33 years; height 1.60 m; weight 56 kg) developed knee flexion-extension movements with the insole, simulating two gait cycles at different velocities to demonstrate the parameter adjustment.

3 Results

Figure 5 shows the variation of M during gait as an example of the application of the method. The footswitch shows a good performance to measure the four sub-phases of the gait. This scheme is implemented in most active orthosis control algorithms, which represents the more common standard for stance sub-phase detection (Taborri et al., 2016). When the user increases the velocity and the footswitch identifies the three phases, as shown in case (b), M maintains a tendency that allows switching the resistance at the knee adequately. In case (c), the gait phase signal shows a noise. During this period, M changes, however, the value follows the correct tendency during the gait cycle. Figure 6 shows the knee angle, M value, knee torque and footswitch signal during a test with the user wearing ALLOR. In IC, MS and TS phases, the torque increases due to the increment of the M value. Then, the knee angle is maintained in 40° until SW phase, when the M value decreases.

4 Discussion and Conclusions

The results presented here indicate that during the walking the footswitch insole allowed the identification of four gait sub-phases. With this information, the method proposed allowed switching the knee block-unblock in the ST and SW phases, respectively, with intermediate sub-phases (IC and TS), in order to obtain a smooth knee impedance variation in the gait cycle. The SC control based on the variation of M and D parameters generates at the user the sensation of walking inside an ambient with high or low damping or viscosity, increasing or decreasing the velocity of the movement and knee torque. The modulation of the admittance parameter M was conducted in volunteers, and the method proposed here can be adapted in cases with gait asymmetries. This allowed the development of a SC that adapts to different impedances at the knee joint in ST and SW phases, allowing a smooth switch. When the user required knee support, higher values of admittance parameters were loaded and reflected knee lock-unlock, however, the gain used to change M and D values depends on the user velocity. In the SW phase, a level of damping to improve an stable movement at low velocities is required, which is different for high velocities. In addition, the gain of the ST phase depends on the user weight, which must be adjusted. Therefore, methods to calculate both the velocity on-line and gain, online, based on user weight, are required, which is part of future works.

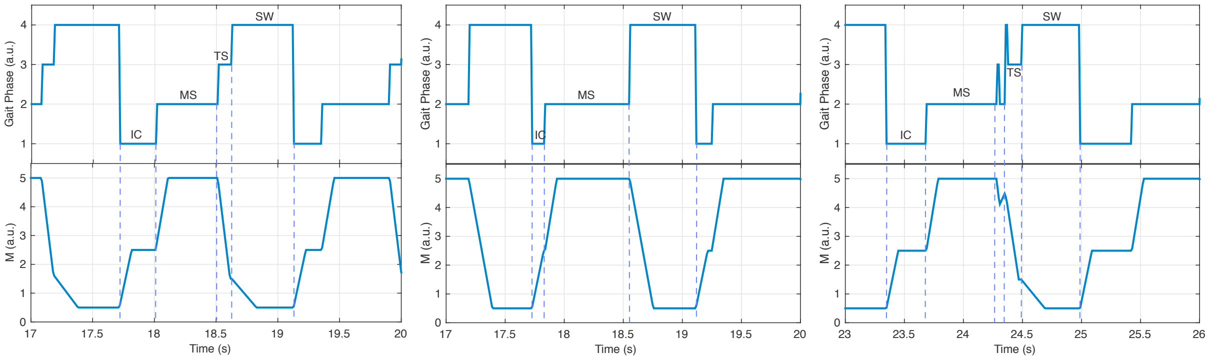

Figure 5: M variation during gait cycle. (a) example of M variation during a gait that generates four sub-phases: initial contact (IC), mid-stance (MS), terminal stance (TS) and swing (SW). (b) example with three sub-phases. (c) example that shows the variation of M during a gait cycle with noise.

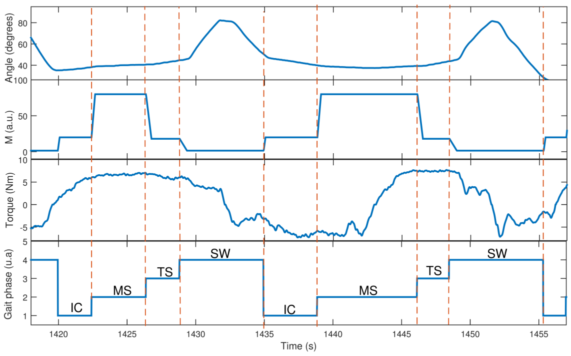

Figure 6: Knee angle, M value, knee torque and footswitch signal during a test with a user using ALLOR orthosis.

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