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# FES-assisted walking with spring brake orthosis: Simulation studies

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**Abstract**. This paper presents a simulation of bipedal locomotion to generate stimulation pulses for activating muscles for paraplegic walking with wheel walker using functional electrical stimulation (FES) with spring brake orthosis (SBO). A new methodology for paraplegic gait, based on exploiting natural dynamics of human gait, is introduced. The work is a first effort towards restoring natural like swing phase in paraplegic gait through a new hybrid orthosis, referred to as spring brake orthosis (SBO). This mechanism simplifies the control task and results in smooth motion and more-natural like trajectory produced by the flexion reflex for gait in spinal cord injured subjects. SBO can eliminate reliance on the withdrawal reflex and foot-ground clearance without extra upper body effort. The stored energy in the spring of SBO is used to replace stimulation pulses in knee flexion and reduce total required torque for the paraplegic walking with wheel walker. The study is carried out with a model of humanoid with wheel walker using the Visual Nastran (Vn4D) dynamic simulation software. Stimulated muscle model of quadriceps is developed for knee extension. Fuzzy logic control (FLC) is developed in Matlab/Simulink to regulate the muscle stimulation pulse-width required to drive FES-assisted walking gait and the computed motion is visualised in graphic animation from Vn4D. The simulation results show that SBO can be successfully used with FES for paraplegic walking with wheel walker with all the advantages discussed over the current hybrid orthoses available.

Keywords: Spring brake orthosis, paraplegic, fuzzy logic control, FES-assisted walking

## 1. Introduction

Paraplegia is impairment in motor and/or sensory function of the lower extremities. It is usually the result of spinal cord injury (SCI) which affects the neural elements of the spinal canal. Sisto et al. [32] reported that more than 200,000 people in the United States (US) suffer from SCI and each year 10,000 new cases occur. Brown-Triolo et al. [3] in their study found that 51% of SCI subjects defined mobility in terms of life impact and autonomy, and gait was found to be perceived as the first choice in possible technology applications. Their subjects also indicated willingness to endure time intensive training and undergo surgery operation if mobility is guaranteed. Therefore, solutions to mobility loss were seen as an exciting prospect to these patients.

Restoring gait in SCI is a research challenge. Researchers have investigated various electrical, mechanical and combined techniques also called hybrid orthosis to restore functional movement in the lower limbs [7, 14, 24–26, 28, 33, 34]. Among the gait phases, the swing phase is important in advancing the leg in order to contribute to movement of the body in the direction of gait progress. Hip flexion is an essential part of pick-up in the swing phase of reciprocal gait, whilst passive hip extension is important during the trunk glide in stance. Researchers have attempted to provide hip flexion to improve walking by a method

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called functional electrical stimulation (FES). FES was first introduced in 1967. It is a technique that uses low level of electrical current to stimulate the physical or bodily functions lost through nervous system impairment, caused by paralysis resulting from SCI, head injury, stroke or other neurological disorders, restoring function in people with disabilities [4]. Currently, applications of FES include standing, walking, cycling, rowing, ambulation, grasping, male sexual assistance, bowel-and-bladder function control and respiratory control. For walking support, all FES-assisted paraplegics need parallel bars, walker or crutches. Moreover, paraplegic walking with only FES has significant drawbacks in function restoration. Firstly, due to stimulated muscle contractions, muscle fatigue will quickly occur because of the reversed recruitment order of the artificially stimulated motoneurons. As a result, there are limitations in standing time and walking distance. Another disadvantage is erratic stepping trajectories because of poor control of joint torque due to withdrawal reflex [11].

Hybrid systems can overcome these limitations by combining FES with the use of a lower limb orthotic brace and might allow paraplegics to ambulate in more natural, efficient manner than they might with traditional passive orthoses. Orthoses can guide the limb and reduce the number of degrees of freedom in order to simplify the control problem. The use of active muscle can also be reduced by locking orthosis joints [9]. Moreover, the approach is useful to support body weight, protect the joint and ligament [21]. Furthermore, its rigidity improves walking efficiency and reduces overall energy cost [31]. Several hybrid systems have been developed. The first hybrid orthosis system combining powered orthosis with FES called hybrid assistive system (HAS) was introduced by Tomovic in 1972 [35]. The work in HAS was continued by Popovic and Schwirtlich [27, 29, 30]. HAS has subsequently been changed to powered orthosis because of use of direct current (DC) motor in the orthosis. Powered orthosis consists of a small DC electric motor installed at one or more joints with or without electrical stimulation support. A functional movement closely mimics the swing phase of gait than the flexion reflex [27–29]. However, this type of hybrid system is not used in practice because of the size and weight of motor and batteries.

The most widely tested orthosis is named reciprocating gait orthosis (RGO) [14, 26, 33]. This mechanism moves the contralateral limb forward by using surface stimulation of hip extension. Then, by alternating stimulation of the hip extensors, walking can be achieved with less energy consumption. However, during the leg-swing phase the body requires to be lifted by the arm with the help of crutches, making it difficult to produce foot clearance. Consequently, muscle fatigue will quickly occur [33].

Goldfarb et al. [9] used controlled-brake orthosis, which is able to address the constraint of FES-aided gait by combining FES with a controllable passive orthosis. This hybrid system includes computer-regulated friction brake at the hip and the knee. Muscle fatigue is reduced by locking the brakes during stance phase and turning off stimulation to the quadriceps muscle. Moreover, leg movement repeats smoothly during the swing phase [9].

Durfee and Rivard [5] introduced energy storage orthosis (ESO) which can be driven through a complete gait cycle. This mechanism uses stimulated muscle power to move the limb and also to drive the orthosis structure, storing energy in the process. Gas springs crossing the hip and knee joints are flexed equilibrium energy-storage elements. The energy store and transfer systems comprise a pneumatic fluid power system connected between knee and hip joints. This can capture the excess energy during the quadriceps stimulation in order to transfer to the hip and release at appropriate instant to achieve hip extension [5].

Kobetic et al. [20] introduced their hybrid orthosis called hybrid neuroprosthesis (HNP). The system uses 16 channels of FES stimulation delivered via chronically indwelling intramuscular electrodes to activate 8 different muscles for the knee, hip and ankle flexion and extension. Electrodes are connected to an external control unit (ECU) temporarily or permanently to an implanted generator powered and controlled via radio frequency by ECU. The variable constraint hip mechanism (VCHM) consisting of hydraulic system with double acting cylinders linked to each hip joint and controlled by energizing specific solenoid valves is designed to maintain hip posture [20]. The result obtained from the clinical test with one paraplegic subject is promising. However, the system size and weight undermine its advantages for the user.

In this paper, a hybrid FES gait system concept called Spring-Brake-Orthosis (SBO) which combines mechanical braces (with coordinated joint locking mechanism) with an energy storage element mounted on it and FES to generate the swing phase of paraplegic gait is presented [8]. This approach also substantially simplifies and reduces the problem of control tasks in a hybrid orthosis while offering more benefits on quality of a swinging leg. Gharooni [8] developed and validated SBO for leg swing phase while Huq [21] used SBO in body weight supported treadmill locomotion in simulation studies. In this paper, the application of SBO is widened where it is used for paraplegic walking with wheel walker. The new concept in hybrid orthotics provides solutions to the problems that affect current hybrid orthosis, including knee and hip flexion without relying on the withdrawal reflex or a powered actuator and foot-ground clearance without extra upper body effort.

Withdrawal reflex is a nociceptive reflex in which a body part is quickly moved away from a painful electrical stimulus. This reflex is due to the signal that does not go to the brain, it loops in the spinal cord and goes straight back from the stimulated area to the muscles that move the leg. Application of FES uses withdrawal reflex for the leg flexion and extension, and is difficult to control accurately. Continuous stimulation will produce delay in withdrawal reflexes, cause more pain and unpleasant situation [2] to the paraplegic. Nakai et al. [23] characterised the flexion withdrawal reflex while Emborg [6] exploited the withdrawal reflex and used the flexes as a feedback to their controller to regulate the swing phase of gait for SCI patients, but it could not be accurately controlled. This is to make sure that the effect of withdrawal reflex on walking gait is considered. Guiraud et al. [10] have eliminated withdrawal reflex in FES by using implantable neuroprosthesis electrodes, but this requires minor operation to insert the electrodes into the subject's limb. In this paper, the use of SBO to overcome the problem associated with withdrawal reflex is presented by using spring for knee flexion to achieve minimum stimulation pulses for knee extension.

## 2. Description of the model

### 2.1. Humanoid model with wheel walker

Humanoid model is built up using anthropometric data. Therefore, the quality and completeness of the anthropometric data are very important in this study. The anthropometric data considered in this study is based on Winter's work [36]. Human body is characterized by three main planes and directions with planes crossing in the centre of the body gravity. The length and mass of each body segment is expressed according to the overall weight and height of the humanoid model. The humanoid model developed in this work is based on a human body with height 1.73 m and weight 80 kg. The locations of the centre of mass together with segment density were also obtained from anthropometric data of Winter [36]. The density of each body segment was used to determine its volume which then determined their segment width.

The wheel walker model is developed using Vn4D software based on the design of a wheel walker sold by Pines Discount Pharmacy$^{TM}$. The model developed incorporated all the basic parts of the real machine. For the wheel walker considered in this paper, the material, dimension and weight are duplicated from real wheel walker that is available in the market [1].

The final stage of the development of the humanoid with wheel walker model incorporated is the combination of both models. It is important to make sure that the humanoid model is attached to the wheel walker model at the right position and right joint. The complete model of the humanoid with wheel walker using Vn4D is shown in Fig. 1.

### 2.2. ANFIS muscle model

In order to simulate FES, a physiological based muscle model is constructed with adaptive neuro-fuzzy inference system (ANFIS) based on previous work [15, 19]. A series of experiments using FES with different

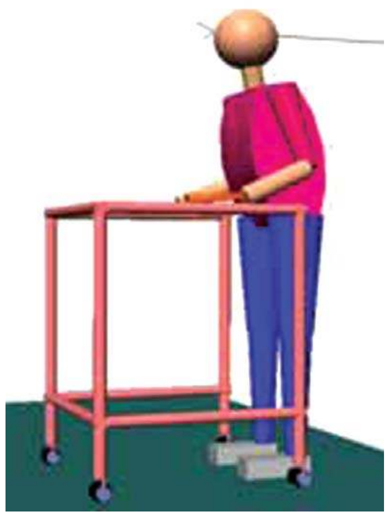


Fig. 1. Humanoid model with wheel walker.

stimulation frequencies, pulse width and pulse duration to investigate the impact on muscle output torque are conducted. The collected data is used to develop the paraplegic muscle model. Five hundred training data and 300 testing data set are used in the development of muscle model. In this paper, the total joint moment generated by the muscle model to drive the walking gait depends on stimulated pulse width as the frequency is fixed to 33 Hz. A fatigue model was incorporated in this muscle model which also takes into consideration that fatigue is increasing with rising stimulation frequencies [16].

In the body-segmental dynamics, total joint moment is the sum of active and passive joint moments. Active joint moment is the addition of the joint moments produced by each muscle group and in this case, it will be represented by muscle model. Passive muscle properties have been separated from the active muscle properties, and are assigned to the joints in order to keep the number of muscle parameters low. In order to make sure the simulation accurately represents the real data, the knee passive properties used in this paper for moment of inertia, damping and stiffness are selected as 0.188 $\mathrm{kgm}^2$, 0.0031055 Nms/deg and 0.024244 Nm/deg respectively [17, 18]. All the parameters used in this paper are extracted from one subject having T2/T3 incomplete lesion for 29 years and results obtained can be further validated by experimental work applied to the particular subject.

## 3. Walking gait

### *3.1. Knee flexion leads to hip flexion*

There are two major forces that act during walking particularly the swing phase; gravity and segment interaction forces. Gravity acts on all masses comprising the body, and for the purpose of analysis, they can all be replaced with a single resultant force acting at the point of centre of mass (CoM). The projection of the CoM on the ground is called centre of gravity (CoG). In the SBO the spring acts as an external force on the knee joint and causes the knee to flex and potential energy is stored in the lower leg (by raising the CoM). Consequently, this causes firstly the shank to accelerate and secondly change in relative angle between the shank and thigh, with the lower extremity taking a new configuration. Both of these produce moments about the hip joint as will be illustrated in the following sections.

#### *3.1.1. Segment interaction*

In the movement of a multiple link mechanical structure such as the arm/forearm system, the torques at the joints arise not only from muscles acting on the joints but also from interactions due to movement of other links. These interaction torques are not present during movement at only a single joint and represent a significantly complicated function in the dynamic analysis of movement [12].

#### *3.1.2. Hip flexion kinetics*

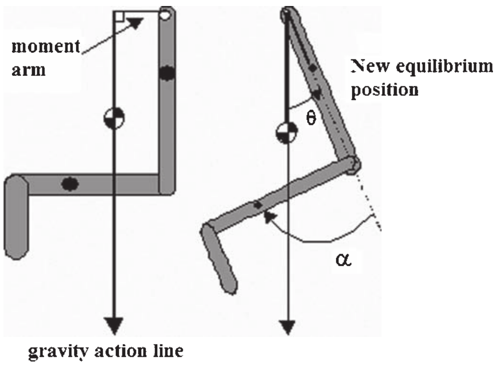
During normal gait, flexion and extension of the hip and knee are linked by bi-articular muscles such as the rectus femoris and the hamstrings group, as well as kinematically and kinetically. Normal gait is initiated by hip flexion with little muscular action around the knee; the inertial properties of the shank cause the knee to flex in response to the accelerating thigh, producing ground clearance [13]. Additionally, as the hip flexes the shank remains in the lowest potential energy position and this leads to additional knee flexion.

These inter-segment linkages also apply when knee flexion occurs without muscular activity at the hip. If the knee is flexed the action of the accelerating shank will cause the hip to flex; additionally, the new orientation of the knee will cause the leg to adopt a new minimum energy configuration with a flexed hip as illustrated in Fig. 2(a). The static relationship between the knee angle ($\alpha$) and hip angle ($\theta$) based on anthropometric data used from [36] is given as:

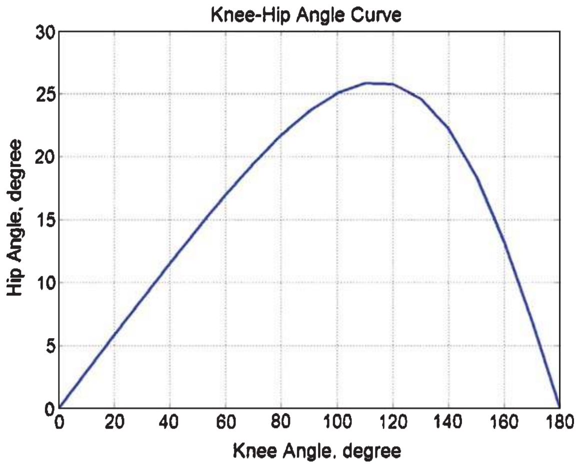
$$\tan\theta = \sin\alpha/(2.426 + \cos\alpha) \tag{1}$$

This relationship is plotted in Fig. 2(b), which represents an ideal situation and assumes no spasticity or muscle contracture. Additional hip flexion is produced by the dynamic inter segment coupling and is dependent on the angular acceleration of the knee. Thus, it can be seen that if the knee can be made to flex by any means then this will also lead to hip flexion. The amount of hip flexion produced by the dynamic inter segment coupling is dependent on the angular acceleration of the knee. Figure 2(c), shows the natural hip flexion produced during knee flexion in SBO prototype developed in this paper. These situations agree with the theory explained in this section.

As indicated earlier, the swing phase is important in advancing the leg and hence movement of the body in the direction of gait progress. During pick-up, hip flexion, knee flexion, and ankle dorsiflexion all combine




(a)




(b)

(c)

Fig. 2. (a) Hip flexion resulting from flexed knee. (b) Static relation between knee and hip flexion angle. (c) Hip flexion angle produced in the knee flexion.

to clear the toe. In this study it is shown that hip flexion can be produced by the knee flexion. Therefore, the only important issue in generating the swing phase is how to produce proper knee flexion. In normal and some FES-assisted walking gait, knee flexion is produced by knee flexor muscle groups such as hamstring. There are two conventional options for producing this knee flexion; direct stimulation of hamstring and use of power actuator. It is possible to directly flex the knee by means of hamstrings stimulation. Disadvantages of this technique are that the hamstrings constitute a biarticular muscle-group which constitutes extending action at the hip and limiting any resulting hip flexion. Hamstrings muscle is also a muscle which easily tends to fatigue. The knee may also be flexed through the use of a powered-actuator such as a DC motor. To minimise inertial properties, it should be mounted away from the knee, as proximal as possible. The previously mentioned disadvantages of size and weight apply.

In this paper, combination of spring and brake at the knee is introduced. The stimulated quadriceps muscles group can usually produce much more torque than is required to extend the leg, even with the thigh horizontal. A spring acts to resist knee extension, then the additional quadriceps torque can be used to 'charge' (store potential energy in) the spring when the leg is extended. A brake can then be used to maintain the knee in extension without further quadriceps contraction, preventing fatigue. When the brake is released the spring will contract, releasing its potential energy as kinetic energy and causing the knee to flex. The advantage of this approach over the use of a powered actuator is that a spring has a very high torque to weight and size ratio, is efficient, robust and does not require any control signals or electrical power. Figure 3 shows the spring for knee flexion and brake used in the developed SBO.

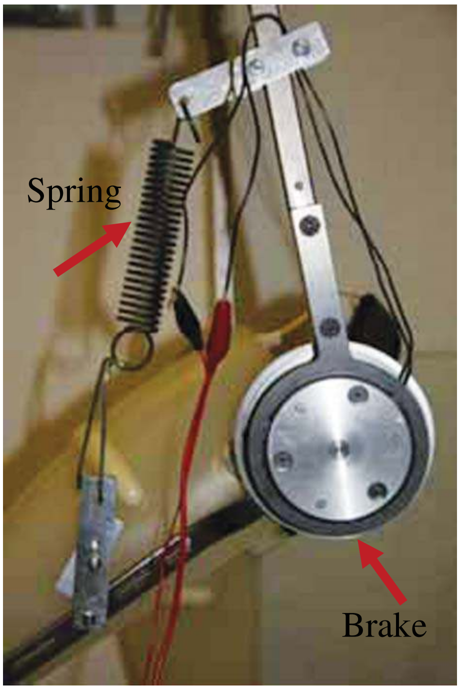



Fig. 3. Spring for knee flexion in SBO.

In order to prevent the dynamic hip flexion produced by the accelerating knee from being lost, a means of 'catching' the hip at its maximum flexion angle is required. This can be achieved by using a ratchet/brake at the hip. This leads to an orthosis combining a ratchet at the hip with a brake and spring at the knee and electrical stimulation of the quadriceps.

*3.1.3. The swing phase with SBO*

Figure 4 demonstrates the swinging leg in the SBO. To synthesise the swing phase of gait using the SBO, the following procedure is required.

1) At the beginning the knee brake is on to provide isometric torque against the spring to keep the leg in stance phase (Fig. 4(a)).
2) The brake at the knee is released, and the spring causes the knee to begin to flex (Fig. 4(b)). It should be noticed that in practice the toe will interfere with the ground at the initiation of swing, and may prevent knee flexion. This problem can be overcome by allowing the unloaded foot to dorsiflex, thus allowing the toe to slide along the ground.
3) Following toe-off, the spring torque will continue to accelerate the shank backwards, producing a reaction at the knee, which accelerates the thigh forwards.
4) The combination of the reaction and the moment due to the weight of the flexed shank cause the hip to continue to flex, the flexed knee allows the toe to clear the ground (Fig. 4(b)).
5) While the hip reaches its maximum flexion angle, the hip ratchet keeps it in peak angle (Fig. 4(b)).
6) The quadriceps muscle is then stimulated to extend the knee against the spring torque (Fig. 4(c)).
7) When the knee is fully extended the brake at the knee is applied and quadriceps stimulation is turned off (Fig. 4(d)).

It can be seen that it is possible to obtain knee flexion, knee extension and hip flexion using only a single channel of stimulation per leg.

## 4. Fuzzy logic control for knee extension

The concept of fuzzy set theory was first proposed by Lotfi Zadeh in 1965. It proposes that humans reason not in terms of discrete symbols and numbers, but in

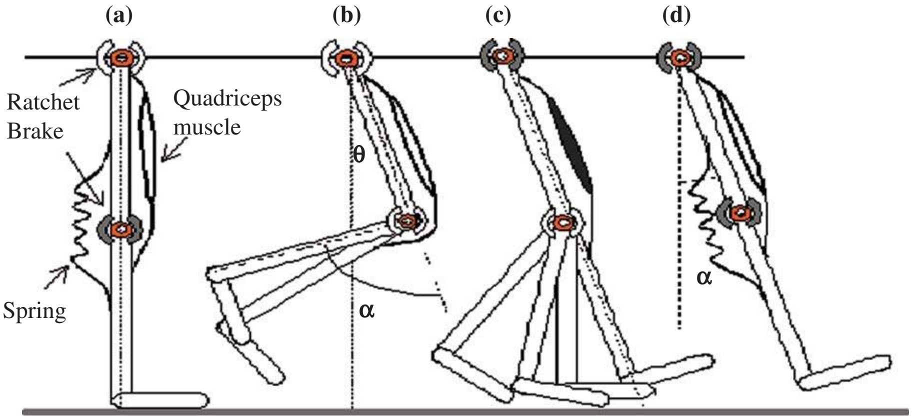



Fig. 4. SBO swing phase synthesis.

terms of fuzzy sets. Today the number of fuzzy logic intervention and projects is enormous and fuzzy logic systems have been employed as powerful tools in many control techniques in different areas such as robotics, medical instrumentation and industry [22].

The essential aim of the fuzzy control in this paper is to make the knee extension follow a pre-defined trajectory by applying a suitable torque to it. Choosing fuzzy controller inputs and outputs is a very critical process, because it is important to be sure that all the information needed about the plant is available through the controller inputs, so as to allow steering the system in the direction needed and be able to achieve high-performance operation. There are 2 inputs selected for the controller. These are the error (difference between actual knee trajectory measured from Vn4D simulation output and reference knee trajectory) and change of error which is the same as the difference between the reference and actual angular velocities. The controller output is the stimulated pulse width which then will be fed into the muscle model to produce muscle torque.

The next stage of the design process is the fuzzification procedure, where the real values of the inputs can be changed to linguistic variables, describing the time-varying fuzzy controller inputs and outputs. Following this process the linguistic values are quantified by using the membership functions. In this controller, five equally distributed Gaussian (bell-shaped) type membership functions are used for each input and output (Fig. 5). Moreover, in this study, equal distribution of the membership functions gives sufficient satisfaction

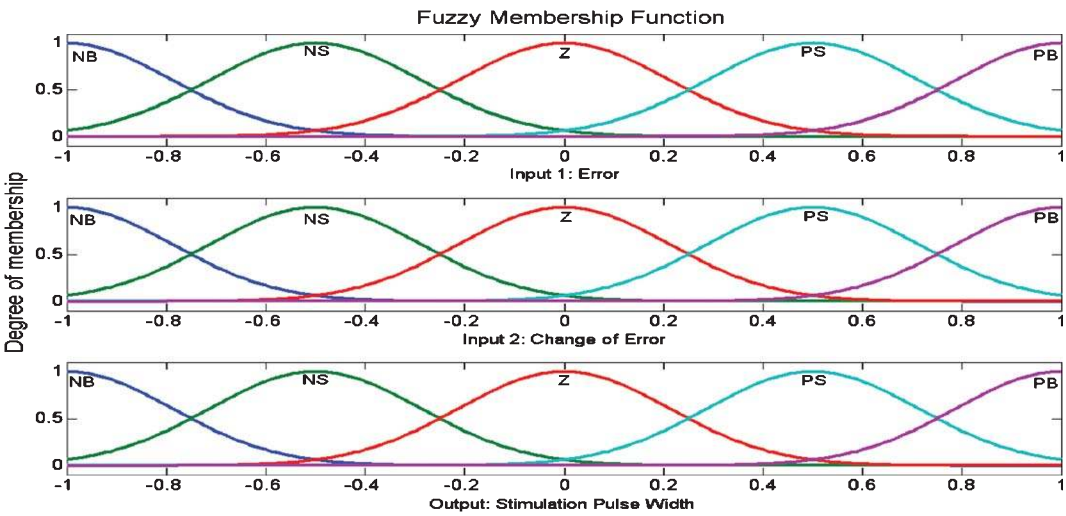



Fig. 5. Fuzzy membership functions.

for the control process, because changing the distribution changes the results significantly. The Gaussian shape for the membership functions is recommended when smoothness is needed.

The next step is to determine which rules to use. The number of rules is determined by the number of membership functions for each input. The fuzzy controller has two inputs with five membership functions and this leads to 25 rules. The determination of the rules is done using the experience of the designer, due to well understanding of the control process, and also tuned by observing the control process, taking important information and building up knowledge on the system. Any combination of two linguistic variables fires at least one rule. They are consistent, with no contradictions and are continuous. Table 1 shows the fuzzy rules for stimulated pulse width for quadriceps muscle. The final step of the design process is the defuzzification method which converts the linguistic values into crisp values by using several defuzzification methods. The centroid of gravity method is adopted because it is commonly used in feedback control due to its smooth output.

The rules are typically fired as:

*If Error is **NB**, and Change of Error is **NB** then the Pulse Width is **NB**.*

The inputs and output are normalised from 0 to 1 and the scaling factor used in fuzzy logic (FL) controller for the left leg are 0.1, 0.0025 and 205 while for the right leg FL controller these are 0.066, 0.0025 and 205 for the error, change of error and output respectively. These were obtained by trial and error process. Figure 6 shows a block diagram of the control system. The stimulation pulse width from the fuzzy controller will feed into the muscle model and produce muscle torque that drives the Vn4D model to follow the walking gait. Then the error and change of error are fed back to the fuzzy controller to adjust stimulation pulse width to the optimum level. In order to apply muscle torque at the correct time, which is the peak time of knee flexion, a block was designed to detect the peak angle of the knee joint. This block sensed the peak time and sent a strobe signal to the controller to initiate controlling the plant.

Table 1
Fuzzy rules for leg extension

| △ee | NB | NS | Z | PS | PB |
|---|---|---|---|---|---|
| NB | NB | NB | NS | NS | Z |
| NS | NB | NS | NS | Z | PS |
| Z | NS | NS | Z | PS | PS |
| PS | NS | Z | PS | PS | PB |
| PB | Z | PS | PS | PB | PB |

## 5. Results and discussion

Simulations were carried out using Matlab/Simulink with incorporation of humanoid with wheel walker model in Vn4D to illustrate the effectiveness of SBO in FES-assisted walking with wheel walker. The Vn4D model is use to represent the subject in the simulation environment with all the parameters precisely obtained to represent the subject as accurate as possible. The control objective is to regulate the level of stimulated pulse width for muscle stimulation in knee extension by following the reference trajectory. The reference trajectory is obtained from Winter [36] referring to the normal human gait based on anthropometric data of one paraplegic subject used in this paper. Therefore, all experiments and simulations are based on the same one subject so that results from this study can be further validated by experimental work applied to the particular subject. The knee trajectories for walking gait and SBO are shown in Figs 7 and 8. Figure 7 shows the result of left knee trajectory while Fig. 8 shows the results of right knee trajectory. Due to various perturbations and limited strength of the hip and knee flexor and extensor muscles, the shank and thigh may not per-

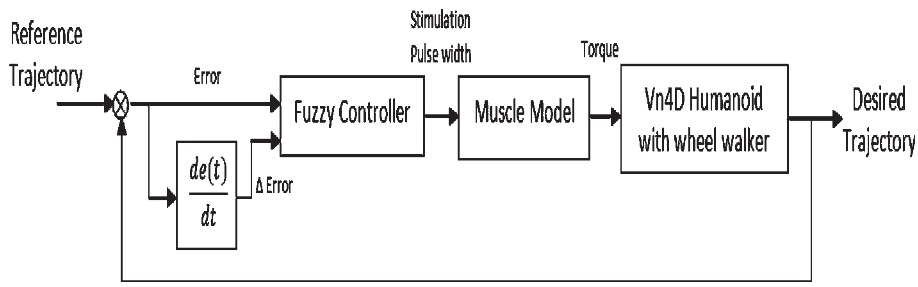



Fig. 6. Block diagram of the control system.

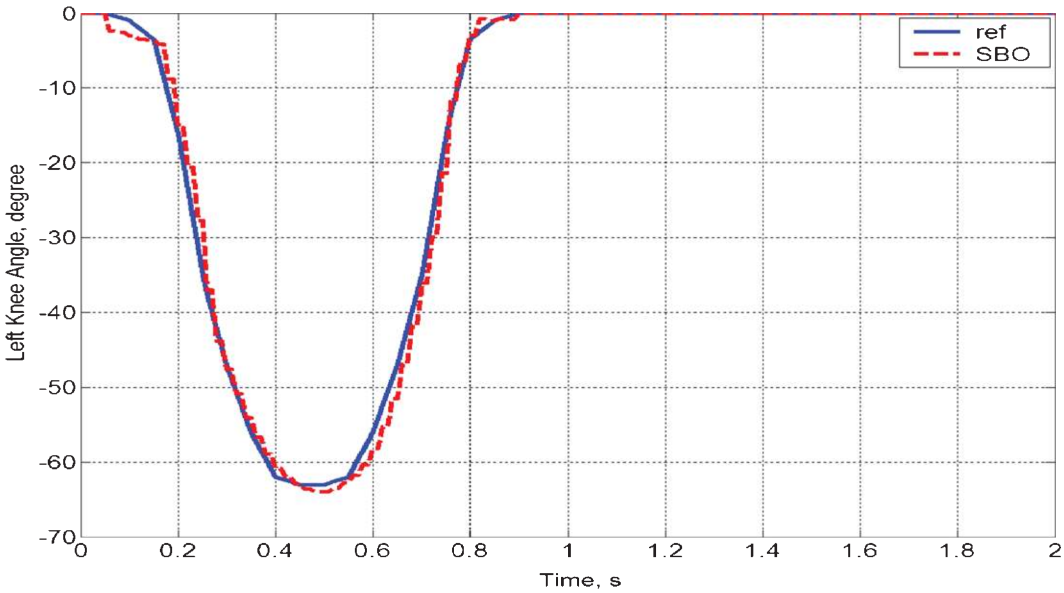



Fig. 7. Left knee trajectory.

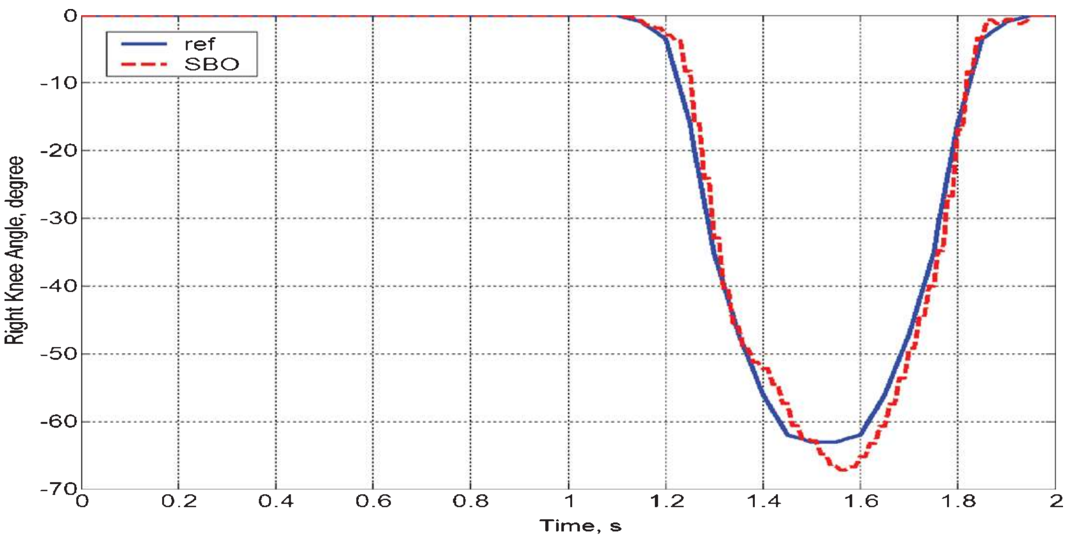



Fig. 8. Right knee trajectory.

fectly track the reference trajectory which is expected when paraplegic is used.

The tasks of swing phase in view of its functional characteristics can be divided into two modes, namely passive and active. In the passive mode, combination of functions of passive elements (brake and spring) initiates a swing phase by flexing the knee joint. A large range of knee flexion picks up the foot to make enough ground clearance. The inertia and pendular effects of the lower extremity advance the leg forward. In this study a spring constant of 200 N/m was used after a trial and error process. This spring constant value gives the best flexion trajectory referring to the predefined trajectory. In active mode, electrically stimulated knee extensor muscles group provides the leg extension so that the heel reaches the ground. The timing block schedules and adjusts the passive and active modes by sending a strobe signal at appropriate times.

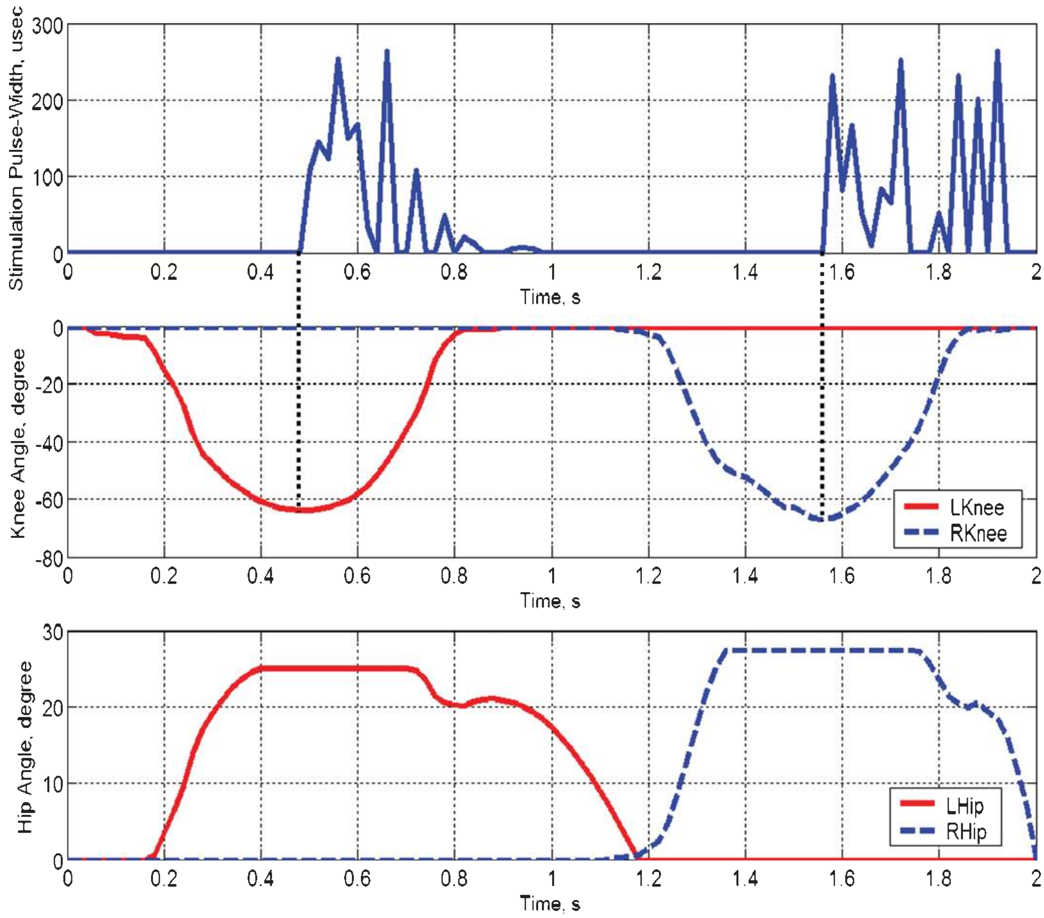



Fig. 9. Stimulation pulse-width, knee and hip trajectory for complete walking gait.

It should be noted that the passive mode is double pendulum driven by only the spring torque in the knee joint and the active mode is a simple pendulum in which the joint trajectory is tracked by the electrically activated muscle torque. This is because the hip brake catches the hip at the maximum flexion angle, and does not allow the hip to move until the end of swing phase.

Figure 9 shows the stimulation pulse width, knee and hip trajectory for both legs. In the left knee, the stimulation starts at 0.48 second and at the time that knee is in full flexion. The same situation takes place with the right knee where the stimulation starts at 1.58 second. The results show that the designed controller works as expected. It is noted that the left hip brake catches the maximum hip angle at 25º while the right hip brake catches the maximum hip angle at 27º. This is because the left leg is beginning to provoke the start of periodic gait cycle and the right leg is where all body parts are in the movement condition and the centre of body is gravity and inertia driven in the direction of progression.

## 6. Conclusion

The objective of the SBO approach is to eliminate reliance on the withdrawal reflex and the associated problems of habituation and poor controllability. Instead, a simple switchable brake with a spring elastic element with well-defined properties provides the necessary function and trajectory. The technique seems promising in producing functional hip flexion. The results of the SBO typical behaviour with the model simulation confirm the effectiveness of the SBO for FES-assisted walking with wheel walker.

The use of Vn4D in this paper is to represent the subject in the simulation environment. All the parameters in Vn4D are precisely obtained using optimisation techniques and experimental data from the subject. It has been demonstrated that FLC can be successfully implemented to regulate the level of stimulation pulse width used to stimulate the knee extensor muscle for FES-assisted walking with wheel walker. Based on the simulation developed, a stable walking gait has been successfully achieved. At this point, simple FLC is sufficient to use in this study, and for future improvement, finite state control can be considered to replace FLC because of its robustness. However, the system validation with the actual system is still pending due to the system set up and clinical approval.

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