Abstract - The objective of the work presented here is to develop a low cost active knee prosthetic devices as real time embedded system which utilizes the available biological motor control circuit property integrated with a Central Pattern Generator (CPG) aided control scheme. The approach is completely different from the existing Active Prosthetic devices, designed primarily as stand alone systems utilizing multiple sensors and embedded rigid control schemes. First we analyzed a fuzzy logic based methodology for offering suitable gait for an amputee, followed by formulating a suitable algorithm for designing a CPG, based on Rayleigh’s oscillator. Using the oscillator we presented a number of simulation results which showed the behavior of knee angles and hip angles and determined the stable limit cycles of the network, and compared them with the captured gaits of an individual. Subsequently, we presented a methodology about how to use CPG outputs for calculating the damping profile for controlling a prosthetic device called AMAL (Adaptive Modular Active Leg).

Index terms: Human gait, CPG, bipedal walking, above knee prosthesis.

I. INTRODUCTION

HUMAN walking is bipedal and due to our inverted pendulum like structure (with comparatively heavy heads) the balancing and controlling of such structure during walking is difficult. This is the reason why, in spite of having an evolutionary advanced brain, a human kid normally takes 8-9 months for learning a rhythmic walking pattern whereas even with a more primitive brain, a quadruple baby calf can walk (and even run) within a few hours of its birth. Learning human walking is a gradual process. Human babies first learn how to balance head & body in sitting posture, then standing posture with and without support, and finally learn balancing in walking cycles. During these initial learning phases, cerebral cortex is deeply involved.

Although the exact mechanism of learning and sharing information between the cerebral cortex and spinal cord is not known yet, it has been proved [1,2] that once the pattern is learnt by the cerebral cortex, the control of repetitively playing back the pattern is taken care of by a group of interconnected neurons (biological motor control circuit) located in the spinal cord, known as CPG and they can be activated without detailed sensory feedback.

In fact CPGs are very sophisticated circuits that can generate complex locomotor behaviors while receiving only input signals. Many models of CPG have been designed for the control of biped locomotion in humanoid robots based on Gentaro Taga’s seminal work on neuromechanical simulations [3]. Some more examples of CPG-controlled biped locomotion include Miyakoshi, Taga, Kuniyoshi, et al [4]; 1998; Shan & Nagashima 2002 [5]; Nakanishi et al 2004[6]; Armando C de Pina Filho et.al 2004 [7]; Endo, Nakanishi, et al 2005 [8]; Righetti & Ijspeert, 2006 [9]; Aoi & Tsuchiya, 2006 [10]. However, although neural oscillators offer simple and robust solutions to the control problems for biped locomotion and dynamic manipulation of robots they are, as such not suitable for human locomotion. The main reason is, the most sophisticated CPGs designed so far for humanoid robots are still not really humanoid. They are not energy efficient. If somebody tries to walk following a humanoid robot’s walking pattern, he/she will waste a huge amount of energy and will get tired soon. Another reason is, that there is a huge variation in humanoid CPGs. For every individual the walking pattern is different and there is not yet a well established design and customization method for mathematically modeling humanoid CPGs. Our proposed methodology is a step towards this. It involves a fuzzy classifier to determine the most suitable gait pattern for an individual which acts as an input for the whole CPG design process. The data supplied by the fuzzy classifier are used for tuning the parameters of an oscillator based on non linear dynamical system. Tuning coupling parameters are of particular importance since here we are interested in utilizing the coupling and residual biological motor control circuits available in the higher joints (in case of above knee prosthesis it is actually hip joints) for an amputee.

In this paper, in section-II, first we have discussed designing a fuzzy logic based classifier for capturing and customizing biological gait patterns followed by developing a CPG design algorithm based on reverse engineering process (in section III). Finally, we have discussed how to use CPG generated trajectories as a reference knee profile for subsequent control of AMAL-like prosthesis (in section IV).
This design paradigm is different from the existing artificial legs available so far to amputees [11, 12, 13]. Existing artificial legs are designed as stand alone systems which are controlled totally from sensory feedback from the artificial leg, requiring various sophisticated sensory devices and various rigid control strategies. This has made the devices very costly, thereby making it beyond the reach of common people especially from the developing world, although a large number of amputees, who have lost their legs due to accidents, diseases, wars or war like situations in the form of sectarian violence etc. live in those countries. Thus the development of low cost, active prosthetic devices is the need of the hour. With the advancement of tremendous storage and retrieval capacity of computers and availability of inexpensive but powerful microcontrollers we can afford to take up this challenge.

We conclude that, since unlike currently available expensive active prosthetic devices, detailed sensory feedback is not required for our control strategy, the design based on our methodology would be energy efficient, inexpensive and would have better acceptability among the amputees.

II. DEVELOPMENT OF FUZZY LOGIC BASED CLASSIFIER

During walking, a normal healthy individual executes the most stable gait patterns. So it makes sense to capture those gait patterns and use them for designing CPG which would subsequently generate required gaits for an amputee at various stages of the gait cycle.

A. Capturing raw data (gait pattern) from various able individuals

In order to capture those gait patterns we have developed a probe called HGD (Human Gait Detector) [14]. The probe uses a rotary potentiometer as a transducer. This, when attached to the joints (knee, hip etc.) of a healthy individual, faithfully generates the voltage time graph corresponding to a particular gait.

B. Developing a fuzzy classifier

The nature of gait pattern of an individual has enough vagueness in terms of variability, inspiring us to try a fuzzy logic based classifier to deal with it. Our objective is to maintain a repository of captured gait cycles for various modes of locomotion like simple walking, brisk walking, running, stair climbing etc. from normal and healthy persons. Using fuzzy logic based classifier a suitable gait pattern could be selected for an amputee who is incapable of producing a normal gait pattern anymore. This will be used for designing a suitable CPG for the amputee.

We have made a fuzzy classifier which at present has 8 classes (depending on height, weight and age). This has been used to predict the closest gait pattern for a person from his height, weight and age. That gait pattern would be used to design a CPG for controlling the damping profile of the amputee.

Our classifier works on the logical assumption that morphologically similar persons can adapt with similar gait pattern. So fuzzy classifier will supply the most suitable gait patterns for an amputee by searching the closet match of a person from the database with that of the amputee and this would subsequently be used for designing CPG for the amputee.

For the present investigation captured gaits have been classified based on age (young or old), height (short or tall) and weight (heavy or light) into eight classes (in the ascending order of the amplitude of oscillation).

The designed HGD has been used to capture the gait patterns of many individuals and maintain a database. For an individual based on his age, height and weight, a fuzzy classifier selects the most suitable pattern which could be adapted using fuzzy operator “AND” and based on the following fuzzy rules.

1. If (age is young) and (height is tall) and (weight is light) then (gait _class is class-1)
2. If (age is young) and (height is tall) and (weight is heavy) then (gait _class is class-2)
3. If (age is young) and (height is short) and (weight is light) then (gait _class is class-3)
4. If (age is young) and (height is short) and (weight is heavy) then (gait _class is class-4)
5. If (age is old) and (height is tall) and (weight is light) then (gait _class is class-5)
6. If (age is old) and (height is tall) and (weight is heavy) then (gait _class is class-6)
7. If (age is old) and (height is short) and (weight is light) then (gait _class is class-7)
8. If (age is old) and (height is short) and (weight is heavy) then (gait _class is class-8)

A typical classifier output using MATLAB has been shown in the following figure:
III. DESIGNING CPG BASED ON RAYLEIGH’S OSCILLATOR

Central Pattern Generators are neural networks capable of producing coordinated patterns of rhythmic activity without any rhythmic inputs from sensory feedback or from higher control centers. They exhibit various rhythmic behaviors in the animal kingdom—both in vertebrate animals and invertebrates.

However, suitable design of system parameters of a CPG through mathematical model for bipedal human beings is a complicated problem since it involves non-linear dynamics. The parameters of these systems are notoriously difficult to tune. We have followed a way which has been described as follows:

For simplicity we have considered three joints for modeling coupling actions. Also from practical point of view these three joints (Hip and two knees) are most significant in terms of generating rhythmic walking pattern. The British mathematical physicist Lord Rayleigh formulated this oscillator way back in the 19th century in conjunction with models of musical instruments. This oscillator can be written in the generic form with the help of following equations:

\[ \ddot{\theta} + c_{ij}(\theta_i - \theta_j) = 0, \ldots \]

where \( \theta_0 \) is the initial state, \( \theta_i \) and \( \theta_j \) are the system parameters of the oscillator to be designed.

Intuitively we have placed the oscillators at the shoulders, hips and knees since these joints have considerable influence to each other in creating walking rhythm. For more accurate analysis one can incorporate the ankle joint in the modeling as well. However, for simplicity and demonstrating the principle, here we have considered three joints (hip and two knees). Solving the set of equations one can derive some analytical relations among the various oscillator system parameters and gait cycle parameters as given below.

\[
\begin{align*}
\eta_i &= \frac{4c_{13}(A_i - A_1) + 4A_i^2\epsilon_1 + A_i^2c_{13}}{12\omega^2 A_i^3}\epsilon_1 \\
\Omega_i &= 2\omega \\
\eta_2 &= \frac{4c_{21}(A_2 - A_1) + 4A_2^2\epsilon_2 + A_2^2c_{23}}{12\omega^2 A_2^3}\epsilon_2 \\
\Omega_2 &= 2\omega \\
\eta_3 &= \frac{4}{3\omega^2 A_3^2} \\
\Omega_3 &= \omega
\end{align*}
\]

For more details one could refer to [7,15]. The captured gait patterns obtained from the Fuzzy classifier (section II), have been used for determining the gait cycle parameters like frequency and amplitude and the following algorithm has been used for finding the design parameters of the oscillators:

Step -1 Read the data files for knee-1, knee-2 and for the hip, coming from the fuzzy classifier.

Step -2 Calculate the parameters \( \Omega_1, A_1 \) directly from the captured gait pattern.

Step -3 Tune the oscillator system parameters:
Step 3.1 Since there is strong coupling between the two knees compared to hip (logical hypothesis) select initially large values for the coupling parameters $c_{1,2} = c_{2,1}$ say, 0.5 compared to hip $c_{1,3} = c_{3,1} = c_{2,3} = c_{3,2} = 0.001$.

Step 3.2 Start with low $\epsilon_i$ values, say 0.005.

Step -4 Calculate system parameters using equation set (2)

Step-5 Solve the equation set (1) and compare its result with the captured gait pattern in terms of period, phase and amplitude of oscillations.

Step -6 If the results match stop and display the system parameters and finalize the oscillator design.

Step -7 Or else go to step 3.2 and try other values of $\epsilon_i$ and then $c_{i,j}$ till the results match.

STOP

A. Simulation results and analysis

From equation (1) and (2) and using experimental initial values the graph shown in fig 3.1-2 is generated by MATLAB programming. It is compared with the captured gait patterns as shown in fig 3.1-1. The closed match confirmed the correct tuning of the CPG parameters. This can then be used for generating knee & hip profiles for the individual for various modes of walking like slow, brisk, walking with long steps etc.

B. Implementation

The implementation includes hardware and software development, integration between them and retrofitting the technology in a professionally manufactured passive leg. Details can be found in [15, 16, 17].

The concept of using minimal sensory information is being implemented in AMAL (Adaptive Modular Active Leg) hardware. In the present model feedback is taken only from hip joint through a rotary potentiometer based probe- one end of which is firmly attached with belt of the subject and the other end with the oscillated shank. The integration among different modules which have been described throughout the paper has been shown in fig 3.2-1.

A. Knee trajectory as input to our controller
B. Control strategy developed for damping
C. Knee moment obtained corresponding to A
D. Damping profile created corresponding to A

Fig 3.2-1 Integration among various modules.

The control strategy of AMAL is conceptually different from all existing prosthetic legs in the sense that it integrates Biological motor control circuits with prosthetic leg control noninvasively. Corresponding to a particular mode of
walk CPG tells what should be the desired profiles of hip and knees and controller decides the corresponding damping profile. Feedback is taken only from the hip joint to check if desired hip profile is executed or not. PD control strategy ensures generation of correct damping profile till the desired hip trajectory is achieved. It enables us to control the system with less sensory information utilizing the biological entrainment from the higher joints. This controller is particularly active in the steady state and remains active for long walking periods. Our controller possesses the ability to deal with any abrupt change in gait by immediately making the damping very high so that the leg becomes locked and temporarily passive to ensure maximum safety to the amputee.

C. Strategy for damping profile generation

Here we have expressed the force to be applied by the damper as a function of the state of the knee (i.e. the knee angle).

As shown in fig 3.2-3, for stable walking, the force applied by the damper, must be maximum when the prosthetic leg is in mid to late stance phase (and the biological leg is in the swing phase). This is because at this state the entire weight of the body is borne by the prosthetic leg and the intact leg is off the ground. The force applied by the damper goes on decreasing, as the hip senses heel contact of the biological leg and the prosthetic leg approaches initial swing, because the weight of the body is getting gradually shifted to the ground. While it swings, then the initial and mid swing require no force from the damper as the main job in these phases is done by the hip muscles, but as the prosthetic leg approached the end of the mid swing then it becomes necessary to apply some force to straighten out the knee again just before heel strike, otherwise the situation may be disastrous causing serious injury and discomfort to the amputee due to loss of balance.

Hence, we see (in fig 3.2-8) a rise in the damping force during terminal swing and this rise in the force continues into the next cycle of the gait. That is why we see the damping force in the initial contact phase starting not from zero but from the point where it had been left at the end of the terminal swing of the previous cycle. The control action has been illustrated in fig 3.2-8 with knee profile as an input and knee moment and damping profile as outputs.

In the figures starting from fig 3.2-4 to fig 3.2-7 we have demonstrated the successful integration between different hardware modules. For example, fig 3.2-4 shows the MATLAB generated knee profile while fig 3.2-5 shows the same as generated from the microcontroller; fig 3.2-6 shows the knee moment profile while fig 3.2-7 shows the same from the microcontroller. Fig 3.2-8 shows corresponding damping profile generated by the microcontroller which is fed into the inbuilt controller of the Magneto-Rheological damper (MR damper) for actuation.
Fig 3.2-7 Microcontroller generates the same (for two gait cycles).

Fig 3.2-8 Microcontroller generated damping profile for two gait cycles, based on our control strategy illustrated in fig 3.2-2.

IV. CONCLUSION AND FUTURE WORK

The paper presents a scheme of developing hardware-software suite for an AMAL like prosthetic devices. It is controlled by minimal use of sensory feedback. In fact only control action initiation and drastic change of walking mode needs some sensory inputs while steady state walking and speed adaptation is aided completely by CPG through biological entrainment governed by next available higher layer biological motor control circuit (which is hip joint for the above knee amputees). The minimal use of sensory feedback makes the system energy efficient and less expensive. While integration among various system modules (fig 3.2-3 to fig 3.2-8) has already been done, future work includes writing and testing exhaustively detailed software codes on the top of the formulated framework and expose the product for field tests with large number of amputees.

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