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Techniques for Dynamic Damping Control in Above Knee Prosthesis

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Abstract

The paper presents a new technique for dynamic damping control based on natural humanoid walk for above knee prosthesis. It has been observed that natural humanoid walking is not solely relied on sensory feed back but also on Central Pattern Generators (CPG) and this CPG produces variation in joint trajectories based on task. Developing task oriented active prosthesis exploiting biologically inspired CPG patterns are of main focus of the paper. Task oriented stable gaits were synthesized using ZMP (Zero Moment Point) approach and the results were compared to kinematics of subjects captured by video streaming. Initial results suggest a feasible solution to gait pattern generation and adjustment of damping profile of prosthetic knee joints to achieve normal humanoid walking.

Key words— Active Prosthetic knee, Biped locomotion, Central Pattern Generator (CPG), Zero Moment Point (ZMP).1. Introduction

1. Inrtoduction

HUMANOID robot research and development of active prosthetic knee joints has made remarkable progress in the two decades.

Commercial availability of humanoid robots such as Asimo(Honda)[1], QRIO (Sony Dream Robot)[2] and HRP -2 (AIST) strengthens the above fact, but still achieving humanoid walking control remains a challenge. The need of a strong support for body weight on stance leg, shock absorption at heel strike and flawless stair climbing pose a lot of problems. Extending humanoid robotic research towards lower limb prosthesis in humans has also taken big leaps with the existence of semi active prosthetic limbs. C-Leg (Ottobock)[3] is an example of such a microprocessor controlled knee joint. Biped locomotion strategies can be used to develop humanoid walking in robots and also in prosthetic limbs. Modern knees are being developed using highly adaptive, electronically controlled hydraulic actuators. These intelligent knees use computational intelligence, and with sensory information adjust damping during the swing and stance phase at different walking speeds [11]. To develop this computational intelligence and obtaining accurate damping parameters is the primary aim of this paper.

Various approaches have been followed in studies around the world for humanoid walking control like passive dynamic walking system (PDW) [4], Central Pattern Generator (CPG)[6][7][8], Zero Moment Point (ZMP) tracking [1][16], learning and divide-and-conquer [10]. Other researchers applied various soft computing techniques to tackle the humanoid walking problem. Such a work is done by C. Paul [5] using Artificial Neural Networks (ANN). Biological similarity and stability of Neural Oscillators for producing sustained oscillations has made oscillator based CPG design a focal point of various humanoid walking studies. A neural system model for 3D adaptive quadruped walking is shown by Kimura et. Al [6] , while a biped walking control has been solved by G. Taga.[7] using oscillator based CPG. A rhythmic arm control using oscillators has been designed by Williamson [8]. Extending these works a design scheme for CPG and its application to active prosthetic knees has been proposed.

Along with walking human beings perform various non conventional gait patterns. We frequently carry backpacks, suitcase, buckets, trays etc. while walking. Stable walking has been realized by various researchers using zero moment point approach [16] but little analysis has been done on weight based gait pattern generation. Miomir Vukobra tovi´c had proposed methods of walking pattern synthesis based on zero moment point (ZMP) [16]. Biped robot stability with the ZMP tracking was also dynamically analyzed by Hirai in ASIMO [1].

This paper attempts to generate the stable gait for humanoid walking with weight using a simplified biped model based on ZMP.

2.CPG based locomotion

2.1. Physical biped model

A 5-link biped robot has been modeled for simulation study of walking robots. Key parameters of the biped are given in table 1.

Single DOF(pitch) is incorporated at hip and knee joints. Such a simulation will help in analysis of human gait as demonstrated later with IGRIP human model simulations.

2.2. Central Pattern Generator

2.2.1 Oscillator mechanism

Oscillator characteristics could be modeled using oscillators developed by Van Der pol , Rayleigh and Matsuoka

TABLE 1
KEY PARAMETERS OF BIPED

	Link 1	Link 2	Link 3	Link 4	Link 5
Length [meters]	0.5	0.6	0.5	0.6	0.5
Mass [Kg]	7.5	5	7.5	5	30

TABLE 2
KEY PARAMTERS OF CPG

	Ta	Tr	b	w	h
Hip	0.05	0.6	2.69	2.0	0.05
Knee	0.1	1.2	2.69	2.0	0.05

. For design of CPG the Neural oscillators developed by

(a) (b)

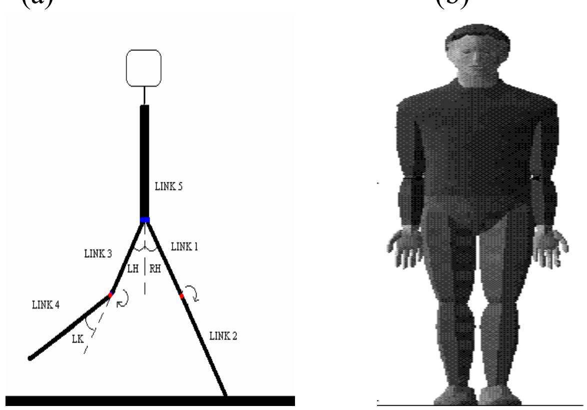


Fig. 1. Bipedal walking models (a) 5-link biped (sagittal view) (b) IGRIP humanoid (frontal view)

Matsuoka [9] have been used which solve the nonlinear dynamics using first order differential equations given below. The stability and entrainment characteristics of the oscillators have been extensively studied and applied in biped and quadruped locomotion [6][7]. The characteristic equations of the mutually inhibiting neurons in a neural oscillator are as follows

$$Tr*\Delta Xi/\Delta t = -Xi - b*Yi + c - w*\max(0,Xj) + \sum a(i,k)*\max(0,X(k))+h*Feed(i) \quad (1)$$

$$Ta\ \Delta Yi/\Delta t = -Yi + \max(0,Xi) \quad (2)$$

$$Tr*\Delta Xj/\Delta t = -Xj - b*Yj + c - w*\max(0,Xi) + \sum a(i,k)*\max(0,X(k))+h*Feed(j) \quad (3)$$

$$Ta\ \Delta Yj/\Delta t = -Yj + \max(0,Xj) \quad (4)$$

Where Tr and Ta represent the rise time and adaptation time for neurons and are responsible for frequency and shape of output of oscillators along with inter oscillator connection weights. X is a variable that represents the degree of fatigue or adaptation in the neuron and b is the parameter that determines the steady-state firing rate for a constant input. The weights(w) of connection between oscillators have been kept constant throughout the simulation. Input c is used to vary amplitude of joint angles.

Oscillatory sensory feedback Feed(i) and Feed(j) provides necessary entrainment with feedback gain h, hence accounting for environment change, which has been shown in Fig 2.

Output of the mutually inhibiting neurons in an oscillator provides the angular change in opposite directions for symmetrical out of phase oscillations.

2.2.2 Oscillator coupling for locomotion

Earlier designs suggesting the existence of two mutually inhibitive neurons for each joint has been followed to generate rhythmic patterns of gait. The oscillator outputs have been used for computing joint angles which correspond to extensor and flexor angles of antagonistic muscles for each joint.

This interconnection of oscillators provides inverse phase relationships between contra lateral and ipsilateral sides for symmetrical motion. Optimal values of weights have been carefully chosen through hand tuning to give acceptable response over a wide range of frequencies and patterns generated for gait.

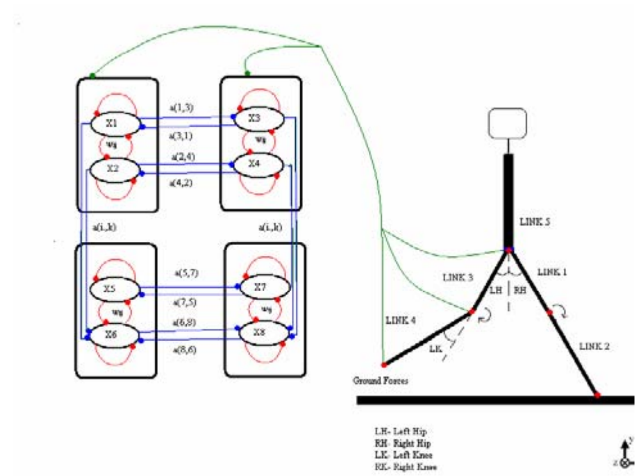


Fig. 2. Interconnections of oscillators and feedback connections.

2.2.3 Sensory Feedback

The importance of feedback is well understood in closed loop systems with feedback providing the interface with the environment for improved control and achieving steady state of the system. Our model also follows the feedback signals for locking of oscillator frequency with the oscillatory input signals from hip angle, knee angle and ground forces as shown in Fig 2.

In human walking visual inputs play an important role in dynamic obstacle avoidance, but small changes in slope and unevenness of terrain can be handled with sensory input from foot only. Paramters used in CPG design using oscillators are given in table 2.

Next section explains ZMP based locomotion and algorithm based decision of most stable trajectory.

3. ZMP based locomotion

3.1. Bipedal Model

The biped planar model used here for analysis was a 7 Link model with 6 Degrees of Freedom given in fig 3(a); model carrying weight is shown in 3(b). The key specifications of bipedal model are given in Table 3. In case of normal walking subject puts the body in Single & Double support Phase [16] alternatively and by applying the forward force during stance it accelerates forward. The time for Single Support Phase is very less (as compare to the Double Support Phase). Also, the Single Support phase is inherently unstable hence a small interval of 0.01 Second is chosen during the Walking Cycle.

3.2. Mathematical Model for Trajectory generation

A model based approach was used for deciding the

(a) (b)

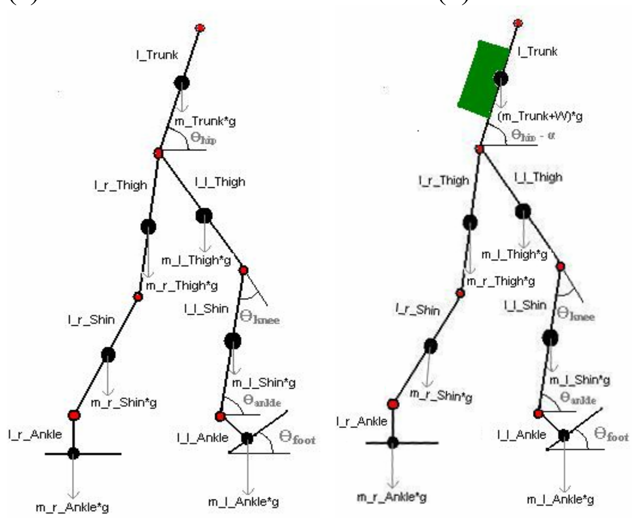


Fig. 3. Sagittal view of Bipedal Model (a) Without Load (b) With Load

TABLE 3
BIPED MODEL PARAMETERS

Link Parameters	Trunk	Thigh	Shin	Ankle	Foot
Mass (In Kg)	43	10	5.7	-	3.3
Length (In meters)	0.50	0.30	0.30	0.10	0.23

trajectories of different joints. The foot and hip trajectory were calculated mathematically and optimized using algorithmic approach given in section 3.3. Finally knee trajectory was interpreted from hip and ankle joint positions.

Parameters given in Table 4 were used for caluculation of foot trajectory through 3rd order Spline Interpolation.

Hip trajectory plays an important role in designing the bipedal walking system and also providing stability to the amputee wearing an artificial limb. Hence an algorithmic

TABLE 4
PARAMETERS FOR WALKING CYCLE

Parameters	Length of Step	Period of Step	Height of Step	Foot Angle (Start)	Foot Angle (End)
Value	0.5 meter	0.9 Sec	0.16 meter	0.2 radians	0.2 radians

approach was devised to obtain optimized walking with varying load on the back of the subject. Hip oscillates from minimum to maximum height in one step of gait cycle. Velocity and acceleration at start and end of each step are assumed to be zero. Several Hip trajectories were generated in a particular posture by varying hip to ankle distances. With the help of ZMP control algorithm we select the trajectory which produces ZMP that covers the largest stability margin.

Smooth and reliable knee trajectory for BKA (Below Knee Amputations) was achieved with the help of the algorithmic technique given in the next section.

3.3 ZMP control algorithm

Algorithm sequence is as follows:

I: Set the biped parameter (e.g. walking speed, obstacle height, step length etc.).

II: Generate the foot trajectory using 3rd order spline.

III:Set the parameters (Hip_Ankle_Forward_Distance (HAFD) & Hip_Ankle_Backward_Distance (HABD)).

IV: If (HAFD > (Step_length)/2 and HABD > (Step_length)/2) goto step V

a: Generate hip trajectory using 3rd order spline.
b: Compute weight based ZMP & Stability Margin.
c: Increase HAFD & HABD.
d: Choose hip trajectory with largest stability margin.
e: exit

V: Exit

ZMP trajectory produced by the algorithm is unique and gives the biped system almost similar movement as subject is having while carrying the weight.

ZMP can be calculated with the help of following formula:

$$Xzmp = \left(\sum_{i=1}^{n} mi(\ddot{z}i+g)xi - \sum_{i=1}^{n} mi\ddot{x}izi - \sum_{i=1}^{n} Iiy\ddot{\Omega}iy\right) / \left(\sum_{i=1}^{n} mi(\ddot{z}i+g)\right)$$
$$Yzmp = \left(\sum_{i=1}^{n} mi(\ddot{z}i+g)yi - \sum_{i=1}^{n} mi\ddot{y}izi - \sum_{i=1}^{n} Iiy\ddot{\Omega}ix\right) / \left(\sum_{i=1}^{n} mi(\ddot{z}i+g)\right) \quad (5)$$

where,

(Xzmp,Yzmp) are the co-ordinates of ZMP position, Linera & Angular Acceleration in Z and X are measured as a parameter in ZMP.

Oscillator based CPG provides the angular patterns in gait cycle and ZMP based synthesis gives a stable trajectory for locomotion. The two approaches are then integrated in the design schema given in next section for damping adjustments in active prosthetic knee.

4. Active Prosthetic Knee control

Knee damping adjustment in semi active prosthetic knee joints is done by trained prosthetists based on the feedback from the amputee. This process is used in C-Leg shown in Fig 4 (b). The damping factors once adjusted remain constant without any consideration of walking speed, weight and gait patterns of a person. Manual interference required to adjust knee resistance is error prone and inaccurate. Hence a dynamic controller is proposed here.

Prosthetic knee control requires efficient sensing (from the view point of effective ness and accuracy) from both global and local sensors which in our design includes knee angle, hip angle and ground reaction forces. These sensed signals are then preprocessed to obtain velocities and accelerations of the knee and hip.

The ground forces are captured using strain gage sensors, although as a simplification of foot model we are using point contact with the ground. The damping given to the knee varies in each phase of the gait cycle depending upon the sensor information. Our CPG model computes the expected angles and velocities depending upon the sensory feedback. Stance phase damping is varied inversely depending upon the position of hip joint with respect to the knee joint. In turn, the swing phase damping is inversely proportional to the difference between maximum flexion angle obtained through CPG and biologically limiting flexion (taken as 70 deg). These computed values are then used to adjust the resistance offered by prosthetic knee joint to achieve biologically realistic gait. The basic design flow of damper controller is given in fig 4. (a).

Biped model was used to give the inputs in the form of

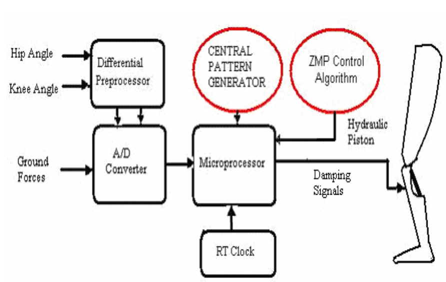


Fig 4.a Active damping control

Fig. 4 (b) C-leg Heel Strike displayed

joint angles and ground forces. The term which primarily affects the gait is the variable weight the subject is carrying values were passed to the ADC module which will serve as an entry gate for the Digital System. ADC outputs the bit stream which was given as an input to microprocessor. The ZMP control algorithm mentioned earlier gives the optimized and stabilized trajectory.

5. Results and discussion

5.1 CPG results

Implementation of CPG and Stick diagram simulations of humanoid biped walking has been done using MATLAB 6.5. The weights of connections between oscillators were decided using hand tuning and remain constant throughout the simulation. Input to oscillators is used to observe the variations in the step length, stride length of biped gait pattern. Time constants have been chosen to receive optimal results close to human like gait and also to vary the frequency of walking.

Joint angles and velocities have also been obtained using human model in IGRIP software.

Leg joint angles vary between the ranges given in table 5. (Measured as in fig. 2)

TABLE 5
ANGLE MINIMUM AND MAXIMUM (IN DEGREES)

Joint	Min	Max
Hip	-30	30 deg
Knee	0	70 deg

The similarity between the oscillator signals and IGRIP simulated gait patterns confirm the correctness of results. Although slight variations from real human gait patterns is seen because of incomplete modeling of foot. First 2-3 steps showed unstable joint patterns which gradually turned into stable gait pattern as desired as evident from the angular plots in fig 5. An increase in frequency of oscillation was seen by giving sensory information to oscillators.

The speed of walking is controlled using step length or varying the frequency of oscillator outputs.

A stable phase response for knee angles were obtained using simulations as shown in fig 7. Although slight varia-

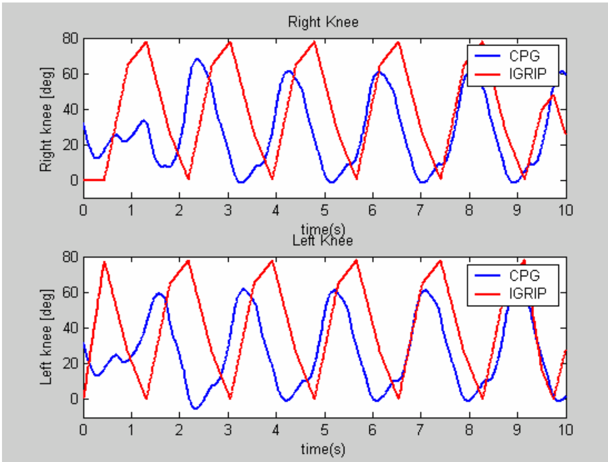


Fig. 5. Angle plot for CPG and IGRIP simulations (Right knee above and Left knee below)

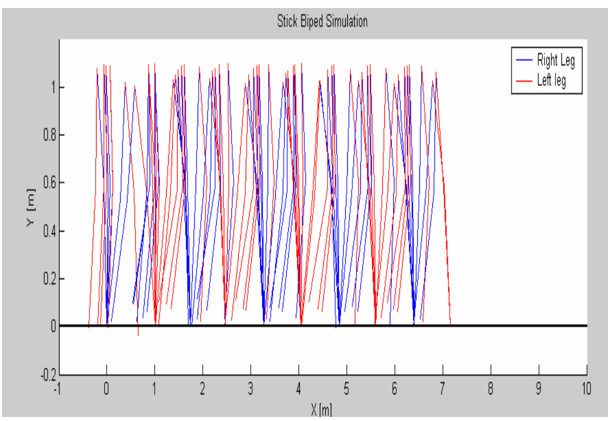


Fig. 6. Stick Picture for Bipedal walking simulations (every 0.4s interval) at c=4

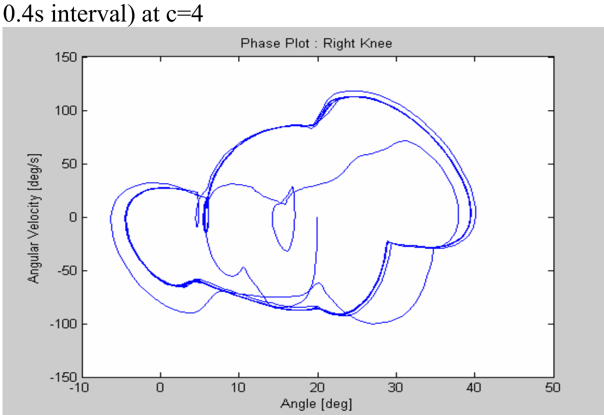


Fig. 7 .Phase plot for Right knee

tions from real walking was seen with the drop of knee angle below 0 degrees.

5.2. ZMP Control

All Simulations were made using MATLAB 6.5. The results were compared to the kinematics of subjects in specific situations. To generate the trajectory from subject's body, the video was captured in sagittal plain and joint markers shown in fig 8 were segmented out through interpolation. For video streaming analysis and segmentation of frames VirtualDubMod was used and ANOVA [18] was used for statistical analysis of variance in Video Data.

Fig. 8. Video frames of gait cycle

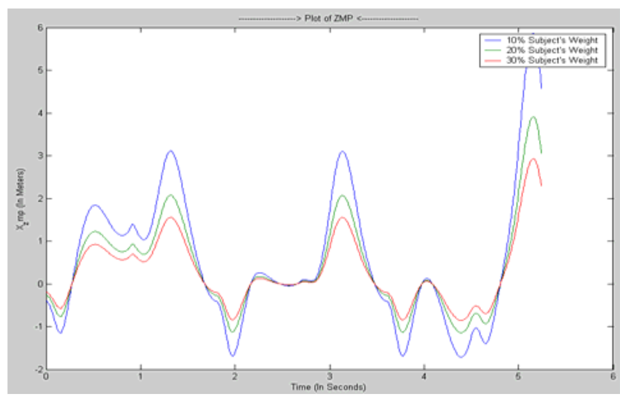


Fig. 9. ZMP Results with varying Load

It was observed that while loading the weight on the back of subject the trajectory of ZMP is shifted down and the step length becomes small as indicated by the plots in fig 9. In critical conditions it has been observed that increasing the weights beyond certain level does not cause much change in trunk bending.

Stick diagram for the Bipedal Model drawn at an interval of 0.1 second is given in fig 10.

Phase plots for hip and knee joint angles is given in Fig.11 above; which demonstrate the stability of angular plots obtained from ZMP approach.

Fig. 12 shows the damping force (which has been defined to be the variable force experienced by both the legs during the different phases of walking. It is measured in Newton) plotted for right and left knee based on control strategy mentioned in section 4. It shows a spike in the middle of stance phase which indicates high damping to resists stance knee bending due to body weight.

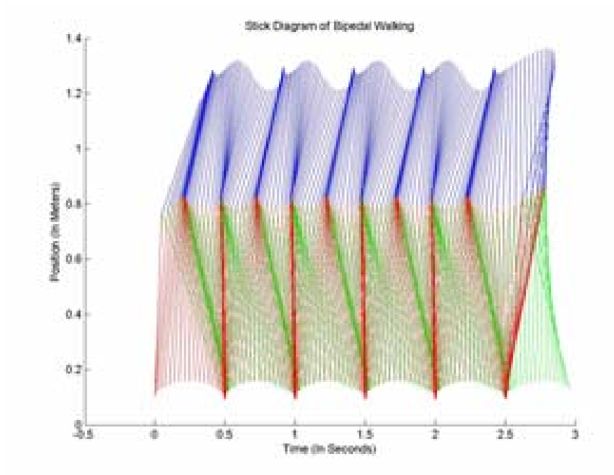


Fig. 10. Stick Diagram for Bipedal Walking (every 0.1s interval)

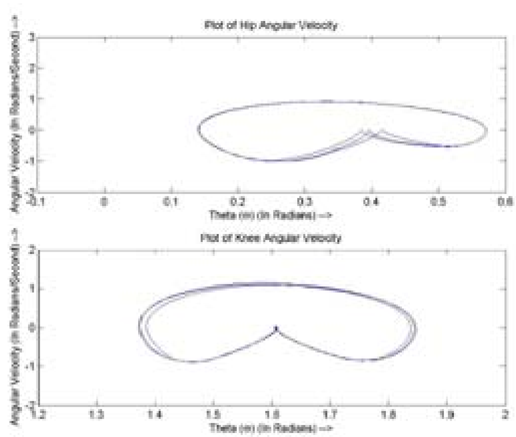
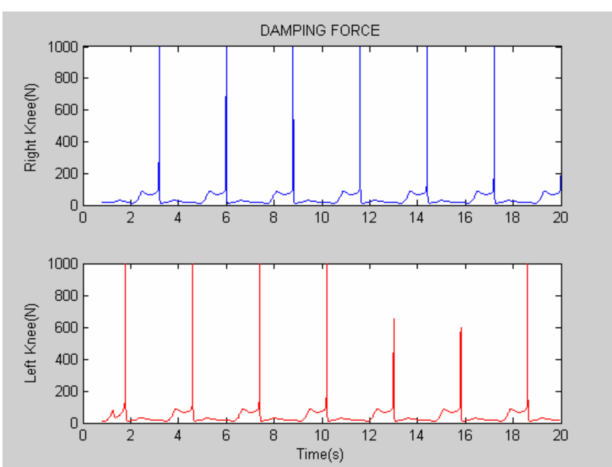


Fig. 11. Angular Velocity of Hip & Knee (in 5 cycles)



6. Conclusions

In this work we verify the results of a rhythm generator based on Matsuoka oscillators to obtain a stable symmetrical pattern of movement between leg joints, and compare it with the patterns obtained by humanoid walking produced using IGRIP. Architecture for evaluation of stance and swing phase damping for active prosthetic knee based on the joint patterns obtained using our simulations has been presented. Unique and stable trajectory was obtained using ZMP for the biped model carrying weight. The trajectory was verified with subject motion through video streaming.
It has been seen from simulations that the gait pattern of an individual significantly depends on the task he/she is performing or the type of load he/she is carrying. So for designing active prosthesis damping cannot be effectively adjusted only based on knee/hip joint trajectories rather the control strategy should include the task based CPG pattern changes to provide maximum comfort to the amputee. It will be an interesting area of research to incorporate this task based CPG in the feedback loop for real time active damping control for a low cost above knee prosthesis.

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