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Conceptual Design of a Powered Ankle-Foot Prosthesis for Walking

with Inversion and Eversion

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Abstract

Human ankle is a series of joints that are highly integrated [1], [2], where tolecular joint (permits dorsiflexion-planter flexion) and subtalar joint (permits inversion-eversion) play vital roles.A conceptual design of an ankle joint is presented here to facilitate the terrain adaptability maintaining natural gait patterns and stability for person with single limb transtibial amputation. The design consists of physiological ankle movements during walking on flat surface as well as uneven terrain. The ankle joint (spherical joint) permits planter flexion-dorsiflexion by the control of passive actuators (springs) and active actuator (motor). It further allows movement in the frontal plane inward/outward about an imaginary centre line of body in order to adapt the roughness of surface. The kinematic behavior of the prosthesis is analyzed. Foot portion of the ankle prosthesis are conceptualize as composite structure to minimize ground contact shock. A 3D prototype is created to represent the conceptual design, which successfully demonstrated the ankles rotational motion.

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1. Introduction

Particularly in India, because of growing urbanization, number of people with lower extremity amputation is increasing due to accident or prevalence of diabetes. Lower limb amputations account for 91.7% of traumatic amputations among which 53% are transtibial amputation (below knee amputations) and 33% are transfemoral amputation (above knee). Before 1980, people having such problem used to wear wooden or metallic structured prosthesis. After that, significant research examining prosthetic foot and ankle components are published. The Seattle Foot (knee, leg, and foot) [3] is the first energy-storing prosthetic component introduced in 1985, that attempted to replicate the natural movement of the foot during various human gaits. The computer control from multiple inputs is introduced in knee prosthesis for better knee stability. These types of passive prostheses can store and return potential energy. But, it is not able to supply the non-conservative positive power or work. These phenomenon attract the researcher further to develop motorized ankle-foot prosthesis.

The full intelligent control of a prosthesis system and adaptation is introduced in 1997 by Ottobock [4], called C-leg. This is a microprocessor-controlled hydraulics knee that adapts the system to all walking speeds in real time. The Proprio Foot (Ossur) [5] is introduced as first bionic or microprocessor controlled prosthetic feet in 2006. Benefits of the microprocessor-controlled ankle movements include the ability to identify slopes and stairs after a few steps, permitting active plantar flexion or dorsiflexion, enabling the prosthetic foot to be flat on the surface for better single limb balance over the prosthetic limb. The powered ankle-foot prosthesis designed by MIT research group Bio-mechatronics [6] is capable of mimicking human ankle behavior in level walking. The active planter flexion-dorsiflexion can decrease an amputees metabolic cost (COT), or the metabolic energy required to transport unit body weight to unit distance, compared to a conventional passive elastic feet. The most recent electrically driven ankle foot prosthesis has been developed by the Arizona State University Human Machine Integration Lab and is called SPARKy3 (Spring Ankle with Regenerative Kinetics) [7]. SPARKy 3 is two degrees of freedom (DOF) device incorporating active control of inversion and eversion, plantar flexion and dorsiflexion. It comprises one additional DOF in frontal plane with active control on ankle inversion and eversion compare to the earlier prosthesis.

From the review related to the development of the ankle foot prosthesis by different researcher, the main challenges may categorized as i) designing a mechanical system which can impart required instantaneous power/torque, ii) finding appropriate bio-signals and their interpretation with gait ambulance, iii) development of control theory for the execution of powered ankle-foot prosthesis [6].

The objective of this work is to develop a low cost ankle for transtibial amputee which will support more natural gait with improved rate of metabolic cost of transport with active control on planter flexion and slightly flexible at side-wise. The deliberations are organized as follows. Biomechanics of human during walking is described in Section II. Section III deals with the design consideration for developing the prototype followed by the mechanical design as described in Section IV and consequent system model is elaborated in Section V. Finally, the results have been discussed towards validation of the design followed by end conclusion as described in Section VI and Section VII respectively.

2. Human Ankle Biomechanics during Normal Walking

Human gait comprises of two phases i.e. stance phase and swing phase through which their normal walking can be done [10]. A stance phase starts with the heel strike and continues till toe-off which contributes around 60-62% of the gait cycle.

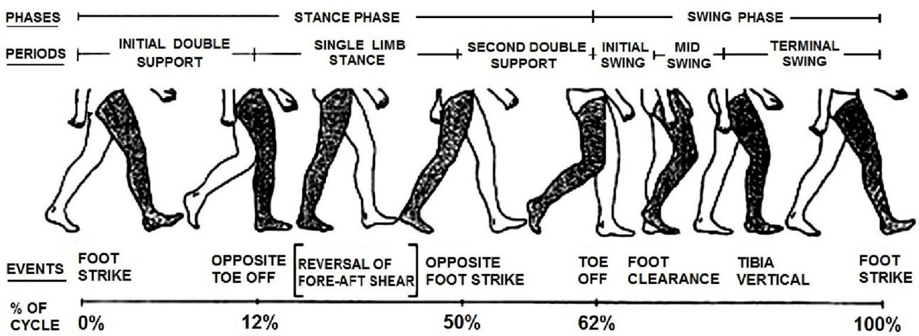


Fig. 1. Human gait during normal walking [11]

Remaining 38-40% of gait cycle is in swing phase. The stance phase can be characterized by the three sub-phases i.e. controlled planter flexion (12% of gait cycle from heel strike to foot flat), controlled dorsiflexion (38% of gait cycle from foot flat to maximum dorsiflexion) and powered plantar flexion (10-12% of gait cycle from maximum dorsiflexion to toe-off) [9]. From the Fig. 1, it is evident that the entire body weight implies on one foot during the single limb stance (12% to 50%) of gait cycle(during remaining phases of gait cycles, the body weight sheared by both the legs or by opposite leg). At the beginning of single limb stance, the CG of body lies straight on the ankle. Then the CG shifts forward (the distance between line of action of CG and ankle increases), which maximizes the moments on the ankle. The moment will be maximumjust before the opposite foot strikes.

3. Design Specification and Targeted Ankle Parameters

From the biomechanical behavior of human ankle, the following salient features are to be achieved in the ankle prosthesis design stage:

1) The prosthesis must give the structural strength to bear the amputee's weight as well as support amputee's daily life activity.
2) The weight of the prosthesis shall be comparable to the missing human limb.
3) The ankle must be capable to deliver required output power and torque during powered plantar flexion.
4) In absence of active power source, the ankle must support passively to continue human gait.
5) The ankle must have lateral flexibility in frontal plane (inversion-eversion) for better terrain adaptability.
6) The prosthesis must provide sufficient shock tolerance for comfort walking and to prevent any mechanical damage to the mechanism.

Conceptualization of the parameters: The target parameters of ankle prosthesis are as follows:

3.1. Size and weight: The size of the prosthesis shall be such that it matches with size of missing biological limb. The height of ankle joint shall be maintained referring amputees other leg (original limb). The height of prosthesis above the ankle joint (that is for housing mechanical components, the actuators, power source, electronics circuits, controllers etc.) shall keep as minimum as possible to facilitate for wide range of amputee. The desired prosthesis mass should be 2.5% of total body mass of the amputee, equal to the percent mass of the missing biological limb at a point 18 cm from the ground surface [6].

3.2. Degree of freedom: The rotational movement in sagittal plane about a single axis can result in good function on level terrain and when using shoes of a particular heel height [2]. But, for walking in uneven terrain or ramped surfaces, it is not always sufficient to have a movement on single plane. Another degree of freedom in frontal plane will improve the terrain adaptation. Therefore, a combination of dorsiflexion planter flexion and inversion-eversion will be an effective means for better terrain adoptability.

3.3. Range of joint rotation:

In sagittal plane: The proposed range of joint rotation in sagittal plane for the prosthesis was based upon normal human ankle range of motion during walking as shown in Table 1. The maximum plantar flexion angle (7 deg.) occurs at the end of stance phase when foot is lifted off the ground, while the maximum dorsiflexion angle (16 deg.) occurs at terminal controlled dorsiflexion [2].

In frontal plane: Inversion and eversion primarily occur at the subtalar joint. Inversion is inward turning of the sole of the foot and the eversion is outward turning [12]. The normal ranges of motion for subtalar inversion are approximately 20 deg. 30 deg., whereas ranges for eversion are between 5deg. and 15deg. [13].

Table 1.Events in human gait during various nature of walking [8].

Nature of walking	Ankle angle (°)			
	Heel Strike	Control Planter Flexion	Max. dorsiflexion	Toe-off
Slow	+2	-7	+16	+3
Normal	+1	-7	+15	-4
Fast	+5	-5	14	-7

3.4. Torque and speed: The peak velocity, torque and power of the human ankle measured during the stance phase of walking gait are as high as 5 rad/s, 1.7 Nm/kg and 3.5 W/kg respectively [9].

3.5. Net positive work: At normal walking speeds, the energy output at the ankle during a single gait cycle is approximately three times greater than the energy output [9]. If walking speed increases, output increases monotonically with relatively constant energy input. Since the energy input to the ankle system is always less than the energy output, it was expected that passive element alone could not characterized ankle function. Active element that could supply energy would have to be included in the model [9]. The ankle foot prosthesis should also be capable of generating net positive work during stance phase. The average net positive work done at the ankle joint per unit body mass for self-selected and fast walking speeds is ~0.10 J/kg and ~0.26 J/kg [9], respectively [6].

4. Mechanical design

Based on the human gait as depicted in Table 1, an attempt has been taken to mimic the walking pattern of an able body by introducing the ankle-foot behavior in the prosthesis. Two basic rotational movements at ankle are adopted in the concept of mechanical design of the ankle. The ankle as a constraint spherical joint can have the rotational motion in two planes within a limit. Putting constraint in one plane, remaining two of a spherical joint may act to provide two rotational movements in one point.

In that way the movement in two different joints (namely tolecular joint and subtalar joint) can be fused into a single joint. Here, concept shows that foot part is connected with adapter through a spherical joint. Salient features are described below.

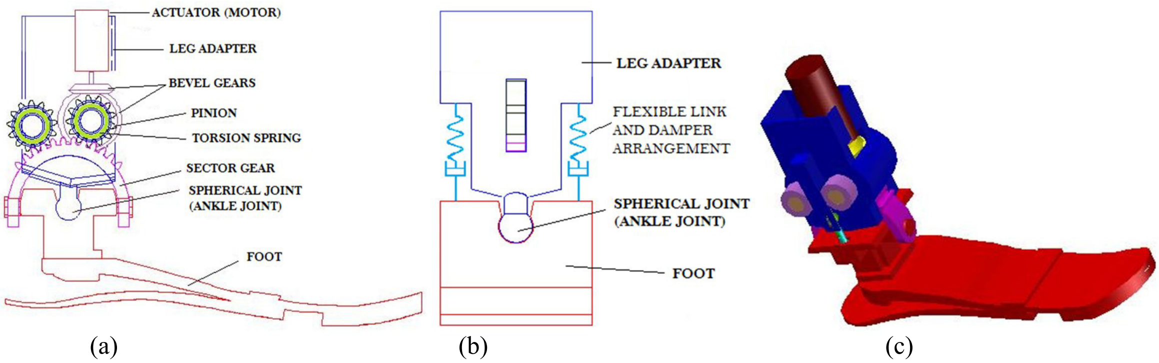


Fig. 2. Conceptual model of prosthetic ankle (a) side view (b) rear view and (c) 3D model

4.1. Transmission of power

As shown in Fig. 2(a), active power would be transmitted through a gear train for the movement in sagittal plane. The rotational movement of gears convert into end movement of sector gear which is pivoted with the foot. The sector gear slides in a curved channel of the leg adapter. With the control of pinion rotation, desired planter flexion dorsiflexion will be achieved during amputees walking gait. Number of pre-stressed torsional springs attached with sector gear would support in flexion and dorsiflexion stage. There is no active power transmission for the case of inversion-eversion. Spring-damper system will provide the necessary flexibility.

4.2. Movement in frontal plane

As already discussed, the lateral movement of foot is required for better adaptability on uneven terrain. During stance phase when foot grounds on rough terrain, it tends to turn inward or outward (eversion-inversion) due to self-weight of the person. For amputee, the twisting action of prosthetic ankle in frontal plane, support more natural gait as well as rough surface walking. This lateral movement of foot prosthesis will be guided by a set of spring-damper arrangement as shown in Fig. 2(b). During walking in uneven surface, a part of ground reaction force due to body weight will be stored in helical springs during stance phase and will be released during swing phase for foot neutralization. As shown in Fig. 3, the damping effect on eversion-inversion of foot prosthesis can be achieved by sliding link mechanism as described later.

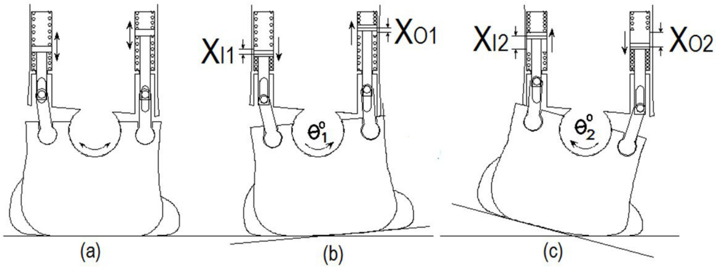


Fig. 3. Passive control of Eversion-Inversion movement (rear view)

4.3. Foot

The foot would include passive low profile Flex-Foot to minimizeground contact shock to the amputee. The "split toe" in foot plate permits frontal motions without absorbing much elastic energy. The advantage is motion with limited loss of dynamic properties.

5. System model

A scheme of powered ankle foot prosthesis is presented in Fig. 4.The net work done at the ankle is broadly in two planes i.e. sagittal plane and frontal plain, while 93% of the work done at the ankle is done in the sagittal plane [9]. However, the majority of ankle work is done in the sagittal plane. Hence, it would be reasonable to consider active control of ankle solely in sagittal plane. It would capture important features of ankle function during stance phase.

As shown in Fig. 2, there are two movements in different planes pivoting in a common point. Around the rotation axis perpendicular to the sagittal plane, a sector gear is moving in a groove by the action of pinions attached with the upper portion. The sector gear is responsible for interior/posterior movement of the foot. Each pinion is driven by a torsion spring andis mounted with sector gear. For position control of the foot with sensory feedback, power driven actuator (motor) and torsional spring are sharing the torque. The ground reaction forces due to amputees body weight, will twist the pre-stressed torsion springs attached with sector gear during controlled dorsiflexion of walking gait. In the flexion stage torsional springs would give positive torque.

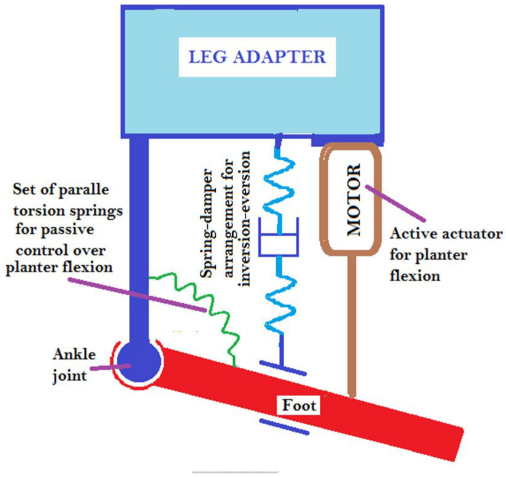


Fig. 4.Schematic of the Powered Foot Prosthesis

5.1. Torque exerted on the ankle

The forces exerted in ankle prosthesis change with the shift of body mass. It considers forces from within the ankle as a person leans as far back and forward as possible while maintaining contact on the ground with the heel as well as the toe. Free body diagrams were constructed to determine the forces acting at different points throughout the ankle while performing activities involved in shifting body mass or standing. Assume that when a person stands on the prosthetic foot, the entire body weight lies on it.

Person standing on the heel of the prosthesis: The weight of the person is laying entirely on the heel as shown in Fig. 5(a) of prosthesis. Therefore, the moment about the ankle joint is:

$$M_{ah} = x_3 B + (x_1 + x_3)m \text{ and } R_H = (B + m)g \tag{1}$$

Person standing on the toe of the prosthesis: The weight of the person is laying entirely on the toe of the prosthesis as shown in Fig. 5(b). Therefore, the moment about the ankle joint is:

$$M_{at} = x_2 B + (x_2 - x_1)m \text{ and } R_T = (B + m)g \tag{2}$$

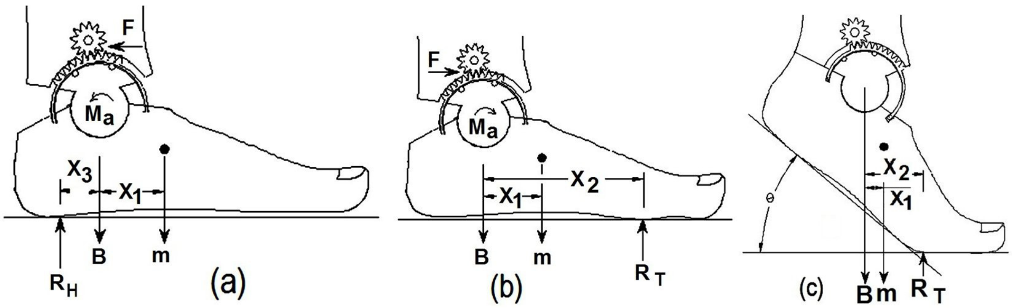
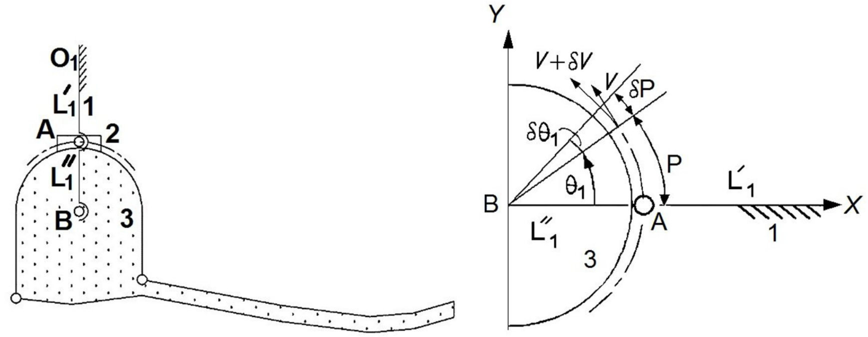


Fig. 5.Forces acting on the prosthesis segment: (a) on heel (b) on toeand (c) on inclination

Peak torque: Comparing above equations (1) and (2), we can say that M_{at} is greater that M_{ah} (because $x_2 >> x_3$ and the mass is negligible compare to body weight). Now, the reaction force R_T is constant unless the body is being loaded by external weight. Weight of foot portion (m) is also constant. Therefore M_a will vary with the variation of $x_2$and x_1.As shown in the Fig. 5(c), as the inclination of foot with ground (θ) is increasing, the horizontal distances i.e. $x_2$and x_1 will decrease. As R_T>>mg, therefore, Ma will decrease with the increase of θ. Hence, maximum torque will be developed just at the time of heel off (i.e. θ is very small).

5.2. Kinematics of the model

Kinematics of the Planter flexion-dorsiflexion mechanism: The Planter flexion-dorsiflexion and related simplified model have been shown in Fig. 6. This mechanism has single degree of freedom. For given rotation, θ_5 at point B, a path, P, will attain at point A i.e. for attending, θ_5 rotation at point B, A will have to slides a distance, P on circular path. $\theta_1$is in clockwise direction from x-axis and is taken as negative.



(a) (b)

Fig. 6.(a) Kinematic diagram of the Planter flexion-dorsiflexion mechanisms (b) Simplified Kinematic diagram

Then, $P = C_3 L_1'' \theta_1 \Rightarrow \theta_1 = P / C_3 L_1''$, where C3 = π/180. The position and velocity of "A" at the instant *t* and (*t* + *δt*) have been indicated. Hence, the change in velocity, δV can be resolved into two mutually perpendicular components δV^n and δV^t. When $\delta t \to 0$, the direction of tangential δV^t component coincides with that of *V*, and normal component δV^n will be directed towards the centre of curvature. The magnitude of velocity at the time t can be expressed as $V = lim(\delta t \to 0)\delta P / \delta t = L_1''\omega_1$. In vector notation, $V = \omega_1 \times L_1''$. The normal and tangential components of acceleration can be derived as follows:

The normal component of acceleration is

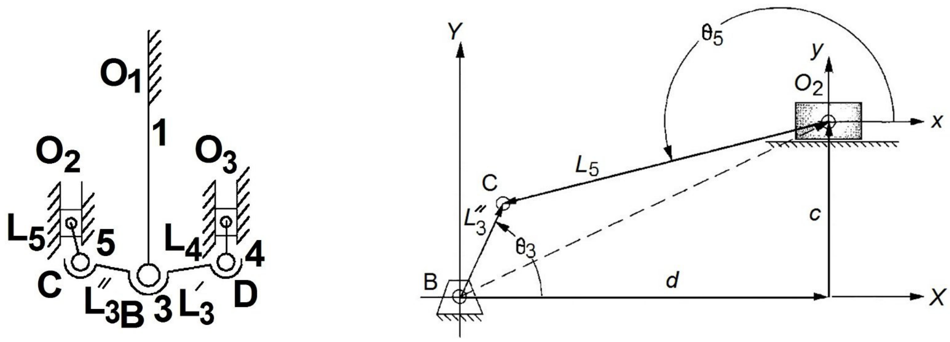
$$a_1^n = lim(\delta t \to 0)\delta V^n / \delta t = L_1''\omega_1^2 . \tag{3}$$

The magnitude of the tangential component of acceleration is

$$a_1^t = lim(\delta t \to 0)\delta V^t / \delta t = (L_1'')'\omega_1 + L''\alpha_1 , \tag{4}$$

where α_1 is the magnitude of the angular acceleration of the radius vector.

Kinematics of the Inversion-Eversion Mechanism: The inversion-eversion mechanism, shown in Fig. 7, is a kinematic mechanism. In this mechanism, there are two sliders connected with two links i.e. L_5 and L_4 respectively. Further, the links L_5 and L_4 are attached with a common link L_3 which is pivoted at point B. For



(a) (b)

Fig. 7.(a) Kinematic diagram of Inversion-Eversion mechanism (b) Simplified Kinematic diagram of the left half

simplification, the entire mechanism can be considered as combination of two similar kind of mechanism. If the left part of diagram is analyzed independently, it is to be noted that the stroke-line of the slider does not pass, shown in Fig. 10, through the axis of rotation of the ankle. θ_5 in the clockwise direction from x-axis is taken as negative.Slider velocity $\dot{d}$,

$$\dot{d} = \frac{L_3''\omega_3 \sin(\theta_5 - \theta_3)}{\cos\theta_5} \tag{5}$$

and the angular velocity of the coupler link ω_5 will be

$$\omega_5 = \frac{L_3''\omega_3 \cos\theta_3}{\cos\theta_5} \tag{6}$$

The acceleration of L_3'', α_3, the parameters such as the linear acceleration of the slider, $\ddot{d}$

$$\ddot{d} = \frac{L_3''\alpha_3 \sin(\theta_5 - \theta_3) - L_3''\omega_3^2 \cos(\theta_5 - \theta_3) - L_5\omega_5^2}{\cos\theta_5} \tag{7}$$

The angular acceleration of the coupler, , α_5 will be

$$\alpha_5 = \frac{L_3''\alpha_3 \cos\theta_3 - L_3''\omega_3^2 \sin\theta_3 - L_5\omega_5^2 \sin\theta_5}{\cos\theta_5} \tag{8}$$

6. Results

The ankle foot prosthesis is proposed here with a set of torsion springs for passive control on planter flexion-dorsiflexion in term to reduce the COT. The eversion-inversion is also guided by a set of damping arrangement for better terrain adaptability. A spherical joint is considered as an ankle joint of prosthesis to have the rotational movements in sagittal plane as well as in frontal plane. The planter flexion-dorsiflexion is guided further by a microprocessor controlled actuator (motor) for active actuation of the prosthesis.

A model as shown in Fig. 2(c) has been developed to analyze the kinetic behavior of the ankle prosthesis. Detail motion analysis has been carried out to validate the proposed concept. Desired motion and their dynamics on application of force in various strategic points considering different terrain condition are analysed and provides satisfactory results.

7. Conclusion and Future Work

This work focuses on the mechanics of the ankle in human walking gait to perform two vital rotational movements. The design is purely mechanical in nature which can attain desired plantar and dorsiflexion angles through active control of actuator. It also permits the lateral movement guided by a set of damping arrangement. The prosthetic foot can act as intelligent device to perform the desired action of amputees' daily life activity. Initially the system is conceptualized as transtibial prosthesis for person with single limb amputation, but proper integration with active knee prosthesis may help to develop transfemoral prosthesis too.

Development of prototype is under progress. Attempt has been taken to generate a Central Pattern by capturing and conditioning (amplification and filtering) of the electrical activity of muscle through EMG. At the same time, attempt has also taken to design a sensor based real time feedback system to determine the most suitable gait pattern properly integrated with CPG using microprocessor based control system enhanced with artificial intelligence.

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