Design of an Artifical Leg Mechanism for Above-Knee Amputees

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ABSTRACT

In conventional designs for above-knee prostheses, there are several shortcomings that lead to energy losses and gait abnormalities for the amputee. Lack of an adequate locking mechanism at the knee and a rigid ankle joint tend to cause defects in amputee gait. This paper discusses a fresh design of a passive, artificial leg mechanism based on gait considerations. A prototype mechanism illustrating the ideas of a weight-actuated locking mechanism and a springy ankle has been fabricated. Preliminary trials have been performed on the device and it is expected that further development in this direction will enable an amputee to have a more natural gait.

Nomenclature

θ_{knee} - Knee angle measured with reference to femur (thigh). (Flexion angle - positive)

θ_{ankle} - Ankle angle measured from the neutral ankle position. (Dorsi-flexion - positive)

θ_{hip} - Hip angle measured with reference to the vertical. (Forward - positive)

W_{foot}, I_{ankle} - Weight of foot & Moment of inertia of the ankle about pin-joint A

G_x, G_y - Horizontal & vertical components of ground reaction force

G,D,d,n - Shear modulus, Spring and wire diameters and number of turns for the spring

1. Introduction

The conventional design for an above-knee prosthetic device involves the use of a simple pivot at the knee, a damper that damps the swing motion of the leg and no articulated ankle joint. There are several shortcomings in such devices, which lead to greater requirement of energy for the amputee. The absence of flexion and extension during the stance phase creates a vaulting action for the amputee, which causes energy losses. Similarly, lack of an articulated ankle also leads to gait defects. For a passive above-knee prosthetic device, the transition of the limb from the stance phase to the swing phase must be smooth. This problem can be addressed by integrating a suitable design for the ankle with the prosthesis.

This paper discusses the design of an artificial leg mechanism for Above-knee amputees. The study of human gait has enabled the identification of certain key features, which when incorporated in a prosthetic knee device, would allow an amputee to walk with a more natural gait. The aim is to design a passive device, which would reduce the amount of energy required. This suggests the use of a controllable damper in conjunction with the device. The configuration is of an inverted slider crank, analogous to the Smart MAGNETIX knee of Lord Corporation [1] and the Ottobock C-leg [2]. The leg mechanism for an above-knee amputee has been developed mainly on gait considerations.

A prototype mechanism illustrating these ideas has been fabricated and is expected to enable an amputee to have a more natural gait. Preliminary trials have been carried out on the leg hardware to test how the locking mechanism engages and disengages during the beginning and end of the stance phase.

2. Key Features of Human Gait

The human gait consists of alternate swing and stance phases [3-4], shown in Fig. 1. The design of the mechanism is based on the assumption that the amputee walks on level ground with a constant velocity of 1 m/s.

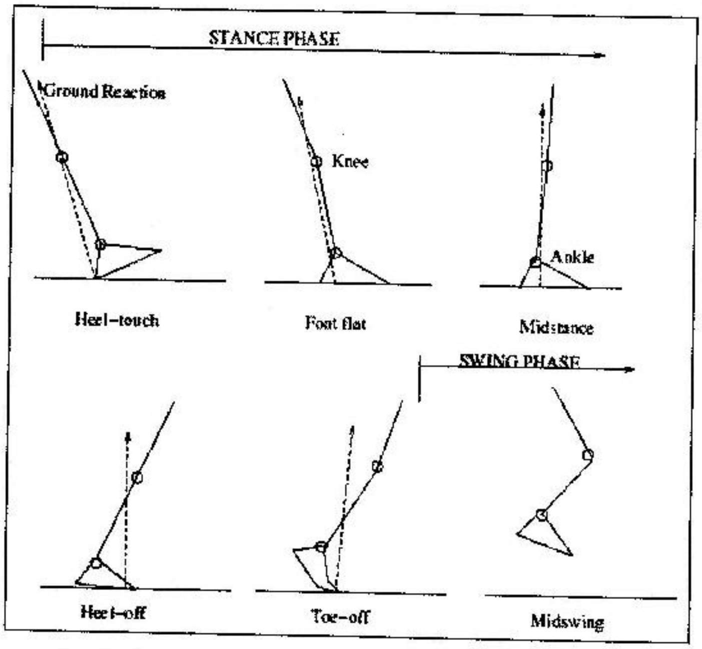


Fig 1: Gait cycle in an average normal human [4].

Some characteristic features of the knee-ankle movements [3-4] are highlighted as under,

1. The knee is fully extended and the ankle is in the neutral position just prior to Heel-touch down. Due to the impact, the line of action of the ground reaction force gets aligned posterior to the ankle joint. This leads to a moment about the ankle, which tends to an undesirable impact between the foot and the ground, termed as *foot-slap.*
2. During Midstance, the knee has to allow flexion up to 10 degrees, to prevent the vaulting effect. But at the same time, the joint should get locked, so that the amputee remains stable.
3. At Heel-off, the knee extends fully and the ankle reaches maximum dorsi-flexion of 15 degrees, as shown in Fig. 2. At Toe-off, the knee flexes through an angle of 40 degrees and it is at this point that the limb requires energy to enable smooth transition into the swing phase.
4. In the swing phase, the knee acts as a damper. The ankle is more or less in its neutral position during this phase.

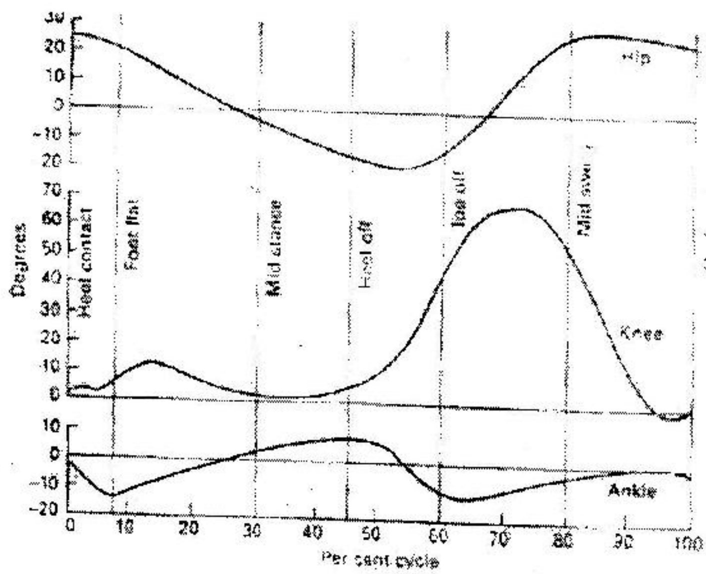


Fig 2: Joint angle variation in human gait [4].

3. Proposed Design for the Ankle joint

An important consideration in a passive prosthesis is to ensure that the limb transitions smoothly into the swing phase. To achieve this, Pratt *et al.* have illustrated the use of a compliant ankle in a 3D biped walking machine [5]. A new idea for a compliant ankle is proposed in this section, which can be integrated with the prosthesis.

A sectional view of the structure of the proposed ankle design is illustrated in Fig 3. It consists of two compression springs. The shank is pivoted at the ankle joint that is modeled as a revolute pair. The neutral position of the ankle is when both the springs are in their natural length and the foot is perpendicular to the direction of the shank. This design serves the following purposes,

- Between *Heel-touch down to Foot-flat,* Spring 1 acts as a cushion to reduce the impact due to *foot-slap.*
- During the stance phase, spring 2 acts as storage of energy as the weight moves over the foot in the forward direction.

3.1 Design issues for Spring 1: This spring comes into action from Heel-strike to Foot-flat. In this section, we derive a relation between the force experienced by the spring and its corresponding deformation. Equilibrium equation is used to compute the moment at the ankle and corresponding force and defomation is computed using geometry of the proposed design. Consider any arbitrary position in that interval as shown in Fig. 4,

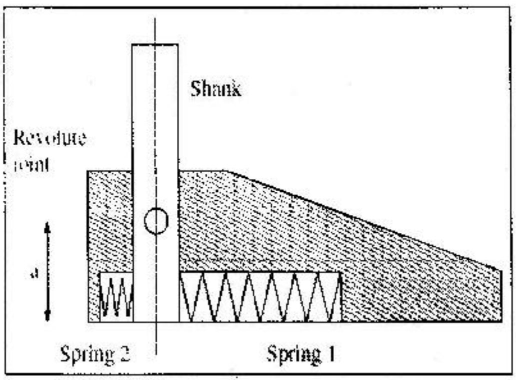


Fig. 3: Proposed design for the Ankle

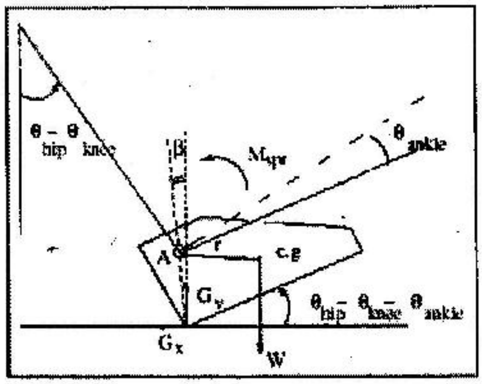


Fig. 4: Moment balance at any point between Heel-touch down and Foot-flat

Moment equilibrium equation about point A is given by,

$$M_{spr} = W_{foot}\, r \cos\left(\tan^{-1}\left(\frac{y_{cg}}{x_{cg}}\right) - \theta_{hip} + \theta_{knee} + \theta_{ankle}\right) - I_{ankle}(\ddot{\theta}_{hip} - \ddot{\theta}_{knee} - \ddot{\theta}_{ankle})$$
$$- l_{AG}(G_y \sin\beta - G_x \cos\beta) \tag{1}$$

Since θ_{ankle} is of the order of 10 degrees (ref. Fig. 2), the relation between force in spring 1 F_{spr}, and the moment it creates M_{spr}, is given by,

$$F_{spr} = ka\theta_{ankle} = \frac{M_{spr}}{a} \tag{2}$$

For five intermediate time-points between heel-touch down and foot-flat, values of θ_{hip}, θ_{knee} and θ_{ankle} are obtained from Fig 2. Values of Ground Reaction force have been obtained from gait data [4]. M_{spr} is computed using Eqn 1 and moment-deflection characteristics are plotted using Eqn 2, in Fig.6 (a). The stiffness of the spring can be determined using the slope of the best-fit line. Using this value of stiffness, the spring is designed using the relation [6],

$$k = \frac{Gd^4}{8D^3 n}$$

3.2 Design issues for spring 2: In order to enable spring 2 to store sufficient energy that can be imparted to the limb to allow transition in the swing phase, it suffices to match the moment variation at the ankle joint due to the spring with that of a natural ankle, during the stance phase. For this problem, at any time in the interval, the moment experienced at the ankle is obtained from gait data. So using geometry of the design, we relate the spring force to the moment and spring deformation to the known ankle angle. Variation of moment experienced at the ankle joint is shown in Fig 5.

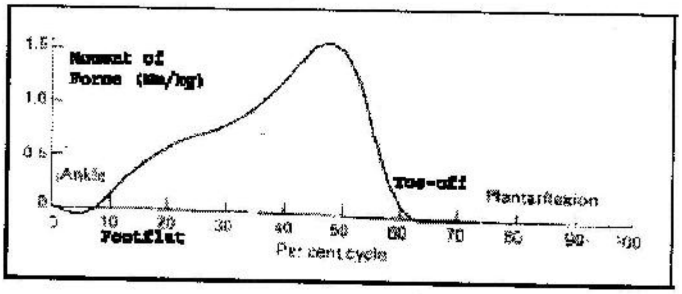


Fig. 5: Torque variation at a natural ankle joint [4]

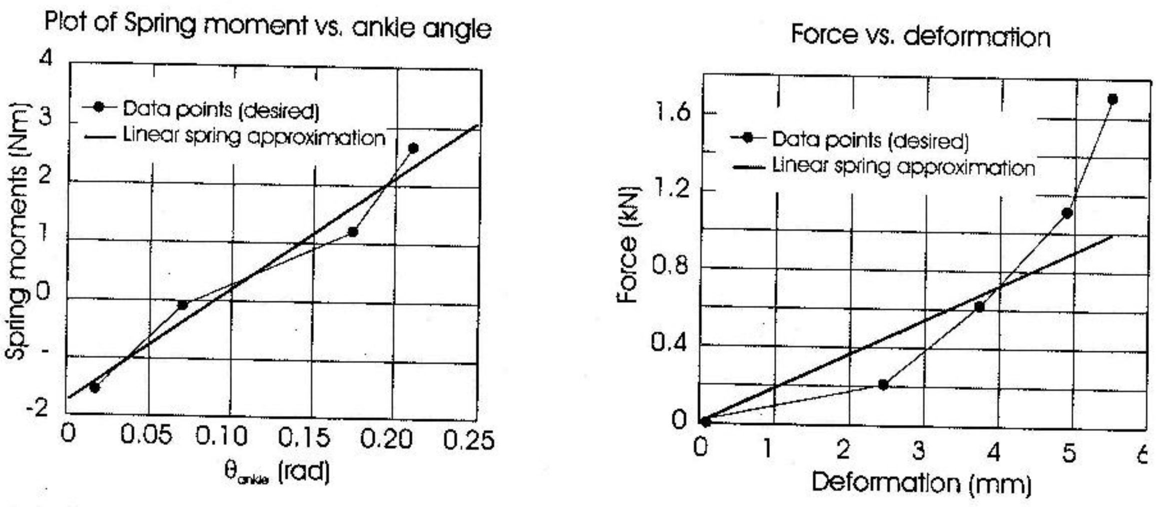
The ankle force is determined at select time-points in the stance phase, using the relation,

$$F_{spr} = \frac{M_{spr}}{a} \tag{4}$$

At the same time-points, the corresponding values of θ_{ankle} are determined from Fig.2. The spring deformation δ is related to the ankle angle θ_{ankle}, by the relation,

$$\delta = a \sin \theta_{ankle} \tag{5}$$

Upon plotting the applied force F_{spr} against the deformation δ at the selected time-points from Mid-stance to Heel-off, a non-linear variation is obtained, as shown in Fig. 6(b). However, the prototype uses a linear spring such that it stores the same amount of energy as the non-linear spring for the maximum ankle dorsi-flexion. With linear spring even though energy stored is matched, the ankle moment profile will no longer match that given in Fig 5. Parameters like spring diameter and free length have been influenced by the space availability in the foot.



(a) Plot of M_{spr} vs θ_{ankle} for Spring 1

(b) Plot of F_{spr} vs δ for Spring 2

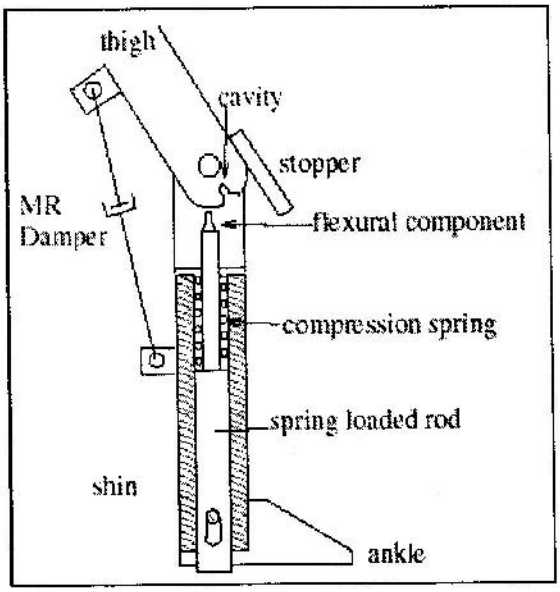
Fig 6 Desired spring characteristics

4. Proposed Design for the Knee joint

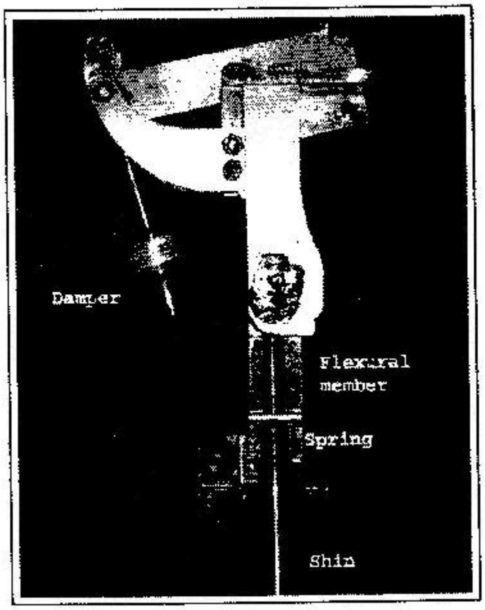
The proposed design for the knee uses an MR-fluid damper in an inverted slider-crank configuration and incorporates a passive, weight-actuated locking mechanism along with the compliant ankle, described in Sec 3. The knee joint needs to be revolute *as soon as the heel lifts off,* while it needs to get *locked* and allows limited flexion immediately after Heel-touch down. A lock has been incorporated that is actuated by the weight of the amputee. The arrangement is shown in Fig. 7. It comprises of a single solid rod that passes through a hollow tube and projects out at the heel base. This rod is spring-loaded and carries a flexural component at its upper tip.

At the beginning of stance phase, when the heel touches the ground, the spring-loaded rod moves through the hollow shin tube due to the weight bearing upon it. The flexural component at the upper tip sinks in the corresponding cavity present on the lower femur surface. They are in line because the knee is fully extended at this point of gait cycle. This arrangement ensures locking and the flexural component is designed as to allow the required flexion during stance.

At Heel-off, the weight stops acting on the spring-loaded rod and the rod comes out of the cavity, due to the compression spring. The knee joint thus becomes revolute. The hyper-extension motion of the knee is prevented by using a simple stop in the opposite direction, similar to the Smart MAGNETIX knee.



(a) The design



(b) The prototype

Fig. 7: The knee concept

4.1 Design Issues related to the Compression spring: The spring used in the rod is of compression-type with a pre-compression of 5 mm. This is to ensure that friction in the motion of the rod through the shin does not prevent release. The spring is designed for 80% of the body weight bearing normally on it. The travel length is restricted to 3 mm. Thus, the stiffness of the spring can be calculated as,

$$k = \frac{W}{\delta + \delta_{precomp}} \tag{6}$$

4.2 Design of the Flexural component: The flexural member is shaped as cantilever with varying cross-section, to induce the non-linear, stiffening effect. The main assumption made here is that *the maximum moment is caused purely by the body-weight* and the resulting maximum flexion is 10 degrees (for body weight of 80 kg), according to Fig 2. A refined shape of the cantilever flexural member obtained by the method of trials using ANSYS is shown in Fig. 8.

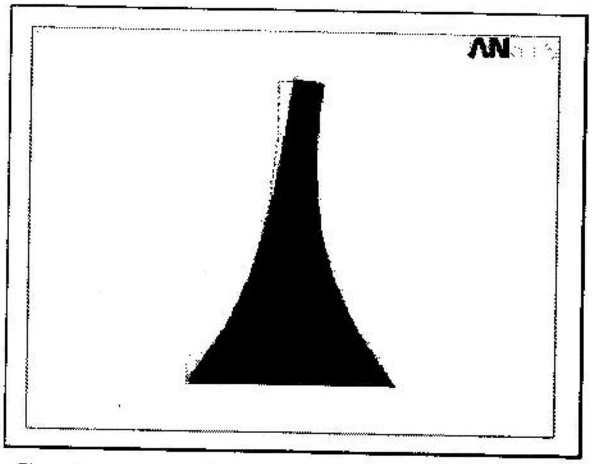


Fig. 8: A refined shape for the flexural portion

5. Results and suggestions for further work

A prototype illustrating the concepts of the knee and ankle design was fabricated and is shown in Fig. 9. Preliminary trials were made on normal subjects with the prosthesis attached to a leg folded at knee. The locking mechanism was found to engage satisfactorily, however, disengaging was found to be sluggish due to friction between the rod and tube. The mechanism worked satisfactorily after a larger diameter tube was used. In manually simulated stance motion, knee flexion of five degrees could be achieved, which suggests that prosthesis may provide the desired flexion during stance for an adult amputee.

The prototype described here demonstrates the feasibility of incorporating several key features in above-knee prostheses. Novel features like weight actuated knee lock that permits flexion and re-extentsion during stance and springy ankle joint tuned for an amputee is expected to enable an above-knee amputee to walk with a more natural gait. However, another prototype needs to be fabricated for conducting the actual amputee trials. Primarily the weight of the prosthesis needs to be reduced substantially before amputee trials can be held. Use of lighter and stronger material, such as aluminium or titanium alloys, carbon fiber composites or plastic would cut down the weight of the prosthesis device.

Shape optimization can be performed to obtain superior designs for the flexural component, which would match the knee flexion angle trajectory in the stance phase. Also, a non-linear spring at ankle joint is desirable to match the ankle moment profile before toe-off in addition to providing energy for lift-off.

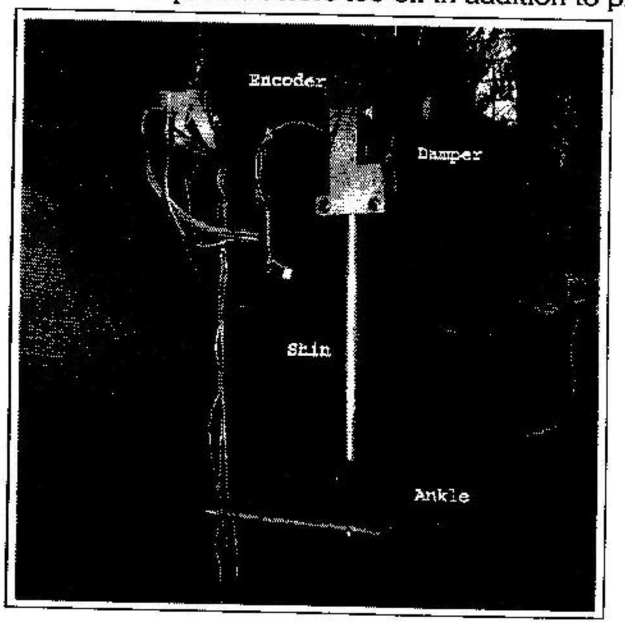


Fig 9: The complete prototype

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