548. Modelling an above knee prosthesis – a kinematics approach

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Abstract. An attempt has been made to mathematically model a totally mechanical passive type pneumatic damper controlled AK prosthesis for swing control. An Alimco, India made AK prosthesis was taken and was upgraded to a pneumatic damper controlled one. A CAD model of the same was made on ProE software and moment of inertia was found and the data were used for calculating the swing time using kinematics model approach. The prosthesis coupled with an electronic interface was experimented to find the swing time. The experimental results matched the theoretical ones within acceptable limits and it is concluded that kinematics model approach as given by the authors is an effective tool to design passive AK prosthesis and the model can also be used for designing active knees.

Key words: AK prosthesis, kinematics model, CAD, swing time.

Introduction

An engineer designing a passive type damper controlled, spring powered above knee prosthesis has to go into complicated mathematics to ensure that the artificial limb swings like a controlled pendulum during extension phase, with speed to ensure comfortable walking which may be slow, medium or fast. The swing time is dependent upon the moment of inertia, the energy gained by an inbuilt spring and the dampers in the mechanism for passive knees. For active knees with motor control, there is need of a mechanism to change the motor speed as walking changes from slow to fast and vice versa. This work gives an insight into kinematic equations which can be used for designing an AK prosthesis for swing time for normal plain walking. For normal gait, the motion of the artificial limb has to match the able limb and for active knees, motor speed and torque are to be selected for swing time control. Faster the speed of walking more is to be the motor speed and for slower walking speed lesser is to be the motor speed.

Prior to World War 2, the above knee amputees used to have a wooden artificial leg and most of the amputees locked the knee joint and continued to walk abnormally. Good amount of work has been done since then, for designing an above knee prosthesis, but the approach used here is innovative and should be of help for R&D personnel working in industry.

For healthy, able-bodied individuals walking patterns are typically symmetrical with respect to kinematic and kinetic variables [1]. On the other hand amputees exhibit asymmetrical temporal-distance (T-D) walking patterns and walk at a slower self-selected walking speed than do able-bodied individuals [2-4].

It has also been noticed that lower-limb amputees take longer steps on the prosthetic side compared with the intact side. They spend more time in stance on the intact leg and in swing on the prosthetic leg. Shorter prosthetic stance has been attributed to amputees as they try to protect their residual limb by spending less time supporting their body weight on the prosthesis for gait stability [4-6]. Gitter, et. al**.** [7], studied relationship between limb mass and mechanical work required for limb movement during swing. It was found that with the addition of 1,34 kg of mass, there was a combined increase in hip flexor muscle concentric work and mechanical energy transfer across the hip joint of 5,4 J, which was needed to accelerate the heavier prosthetic limb into the swing phase. Rietman, et. al. [8]**,** performed extensive review of the literature from 1990–2000 with the topic: “clinical use of instrumented gait analysis in patients wearing a prosthesis of the lower limb”. Zahedi et. al. [9], described the method of defining and measuring the alignment of lower-limb prostheses and quantitatively established values for bench alignment position and the range of adjustment required for incorporation into the design of new alignment units. It was established that an amputee can tolerate several alignments ranging in some parameters by as much as 148 mm in shifts and 17 degrees in tilts. Gitter, et. al. [10], studied and emphasised that the effect of altered center-of-mass (COM) kinematics and increased mechanical work account for the adverse effect on oxygen consumption. The muscle work needed to complete a stride was significantly greater in normal subjects compared with amputees. These results indicated that the role of abnormal kinematics and their effect on the mechanical work of walking are complex and incompletely understood. Orendruff, et. al. [11], tested the hypothesis of microprocessor-controlled prosthetic knees, which claimed to improve gait efficiency in trans-femoral (TF) amputees. This hypothesis was tested in a prospective randomized crossover trial that compared the Mauch SNS knee and the C-Leg microprocessor-controlled knee in eight TF amputees. The C-Leg caused small reductions in net oxygen cost that were not statistically significant compared with the Mauch SNS at any of the walking speeds ($p > 0.190$). Jeffrey, et al., [12] examined the role of the knee controller in relation to a SACH (Solid Ankle Cushioned Heel) foot during the stance phase of level walking. L. Yang, et. al. [13], investigated the effects of angular motion constraints of the lower limbs on the gait of normal subjects. John P. Paul [14], stated that throughout the history of development of joint replacement implants and external prostheses there have been mechanical failures due to discrepancy between material strength, cross-sectional characteristics and the loads developed in normal or abnormal function by the patient utilising the device. Prostheses there have mechanical failures due to discrepancy between material strength, cross-sectional characteristics and the loads developed in normal or abnormal function by the patient utilising the device. Natalie Vanicek et. al. [15], designed and developed a prosthetic simulator to allow able-bodied individuals to walk in a novel situation and studied the changes in gait variables during a novel walking task when lower-limb mechanics were altered and generated good amount of kinematic data. T. S. Bae et. al. [16], evaluated quantitatively the contribution of each muscle to amputee gait during level walking and stair climbing in transfemoral amputees and healthy individuals and concluded that time – distance and kinematic parameters by gait analysis showed no statistical difference between each task, excluding pelvic tilt, pelvic obliquity and hip abduction.

Before fabricating a model or a prototype, engineering approach is to do a mathematical modelling and thereafter design and fabricate the prototype. This work gives an approach of using kinematics equations to mathematically model an AK prosthesis.

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Methods

2.1 Mathematical Modelling

The swinging part of an above knee prosthesis is a compound pendulum, which is constrained from going into hyper-extension. The system as represented through Figure 1 has a damper, a spring and mass. Torque responsible for swing is given by relation:

$$\text{Torque} = mg \sin\theta * b + f(x) \tan\theta * x1 \qquad (1)$$

In equation 1, mg is the gravitational force of the swinging part; b & x1 – are the distances from pivot to centre of gravity of assembly and to cylinder joint at bottom at maximum flexion respectively and f(x) represents spring force. A second order differential equation as mentioned in equation 2 also represents the system.

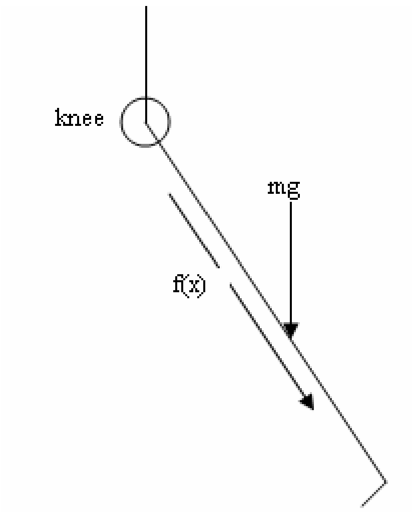

Fig. 1. A line sketch showing the forces acting on spring loaded passive type AK prosthesis

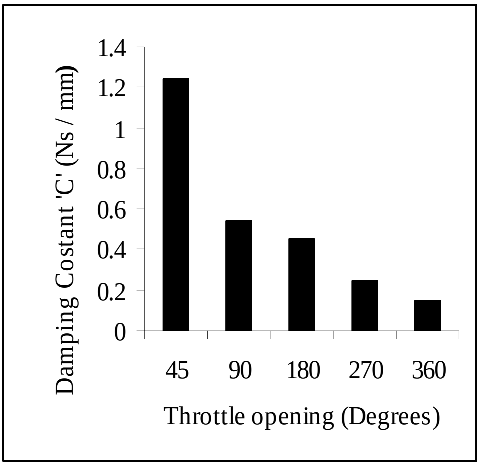

Fig. 2. A graph between damper constant and throttle opening for the hardware used in the experiment. As the throttle is opened damper constant reduces

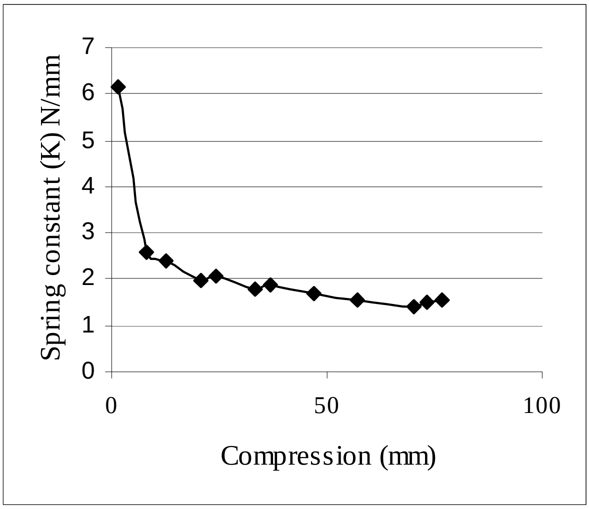

Fig. 3. Graph between spring constant and the compression for the spring selected in the hardware. Being a non-linear spring, the spring constant depends upon compression at a particular flexion angle

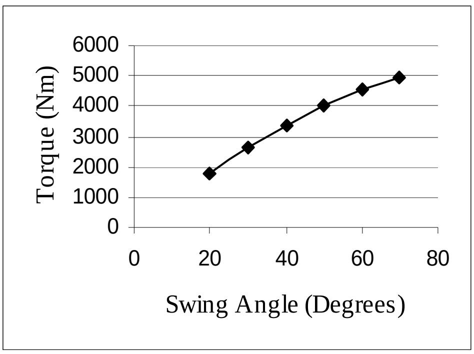

Fig. 4. The graph is clearly showing that torque available to cause swing of the prosthesis is more for more swing angle, and is non-linear pattern

$$\text{Torque} = I\, d^2\theta/dt + C\, d\theta/dt + K\theta \qquad (2)$$

In equation 2, I is the moment of inertia, C is the damper coefficient, and K is the spring constant. The damper chosen was experimented to find C value at different opening of the valve. The value of C has been graphed in Figure 2.

The non-linear spring used in the prototype was experimented to find (a) its spring constant in compression and (b) compression at different flexion angles. Torque responsible for swing at these flexion angles was calculated as per Equation 1 and the torque available from 20 degrees to 70 degrees flexion angles is shown in figure 4. Value of mass was taken as 1,67 kg.

2.2 CAD Modelling

Finding centre of gravity and moment of inertia is a tedious process and CAD softwares do the job easily. The centre of gravity and b value in equation 1, moment of inertia in equation 2 were found using CAD model analysis using ProE software. From the software, Distance of COG from lower axis of rotation (b) came to be 277 mm; Moment of inertia I at COG is 45,33 tonne mm ^{2}. So, I at axis of rotation is 173 tonne mm^2.

2.3 Kinematic Modelling

From kinematics of rigid bodies, following formulae were used:
(1) $\omega = \omega o + \alpha t$;
(2) $\theta = \omega o\, t + 0{,}5\, \alpha t^2$;
(3) $\omega^2 = \omega o^2 + 2\alpha\theta$,
where, ω is the angular velocity, t is time, θ is angular displacement, α is angular acceleration. At t =0, ωo was taken as 0 & θ varies from 20 to 70 degrees. All calculations were done in N, mm, sec units, and K value of shaft was taken to be zero.

The equations were solved as under:
$\omega o = 0$, at t = 0; & θ varies from 20 to 70 degrees.

Our case is one with a damper attached to the prosthesis. The damper used is a linear one, but is restricting rotational motion. Considering that damping value remains same and no rotational spring effect is there, the calculations were done for 20 degree rotation to 70 degrees. One equation for 20 degrees flexion is detailed as under.

The torque required to move the leg now becomes:

$$T = I\, \alpha + c\omega,$$

$$1788080 = 217000 * \alpha + 0.143 * \omega. \quad (3)$$

Further,

$$\omega = 0 + \alpha * t, \quad (4)$$

$$\omega^2 = 2\, \alpha\, \theta. \quad (5)$$

Equation 3 becomes $\alpha = \omega / t$.
From equation 4 $\alpha = \omega^2 / (2 * .3478)$.
Thus, $\omega / t = \omega^2 / .6977$; $\omega = 0.6977 / t$.
Putting the value of ω in equation 3:

$$1788080 = 217000 * 0.6977 / t^2 + 0.143 * 0.6977 / t. \quad (6)$$

Taking 1 / t = x, we get: $151400.9\, x^2 + 0.015345\, x - 1788080 = 0$.
Similarly more equations were derived.

2.4 Experimental Model

A locally available prosthesis of Alimco, India made was taken and modified to make it a damper controlled one. An ATMEL micro-controller based electronic circuit was designed to find the swing time. The prototype was then fitted to a stand, and a microcontroller based circuit powered by a battery was fitted as is shown in figure 5. Swing time for different flexion angles at varying damper coefficient was measured.

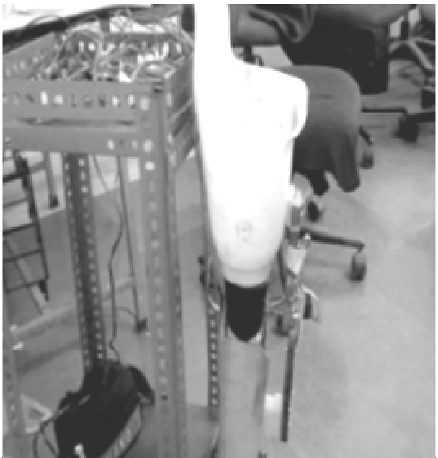

Fig. 5. The prototype of prosthesis fitted to a stand and a data acquisition system

Results

The swing time of the prosthesis is to vary with change in speed of walking and the damper control plays an important role. The check valve opening of the damper fitted in cylinder controlled the damping in the pneumatic prosthesis. The valve opening was done from 45 degrees to 360 degrees, and C values were experimentally found as shown in Figure 2. After 360 degrees of damper opening, no changes were observed. This gave the operating range of the damper used.

The spring selected is a non linear one and the spring constant depends upon compression. The data was used for generating the mathematical equations to find the swing time. Compression spring constant is clearly graphed in figure 3 and shows its non linear behaviour.

For a particular flexion angle, swing time was found for varying damper valve opening. Figure 6 shows the calculated values. There is very small variation but it is not evident, an enlarged view will show these changes but variation with change in flexion angle is phenomenal. Actual condition results shown in Figure 7, clearly show that variation is there up to 300 degrees damper valve opening. Figure 8 compares the mathematical model with the actual swing time for a 90 degree damper opening. The error though is less, and starts reducing as the swing angle is increased and finally converges for 70 degree flexion.

Figure 9 compares the mathematical model with the actual swing time for a 45 degree damper opening. Here error is still less and the mathematical model is almost representative of the actual case.

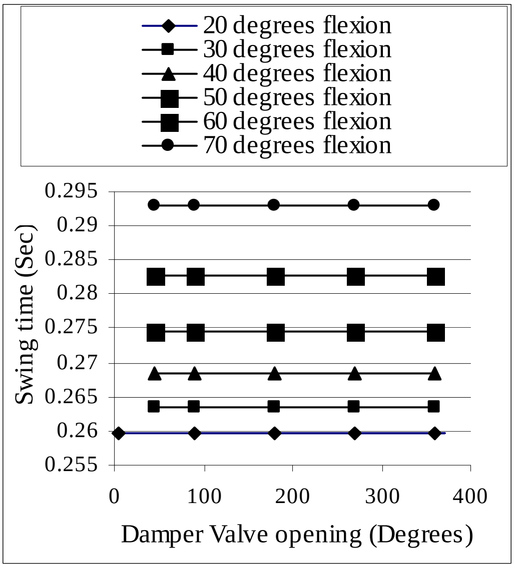

Fig. 6. The mathematical calculations show that with change in flexion angle swinging time increases, but change in damper valve opening is not having a major effect on swing time for a particular flexion angle

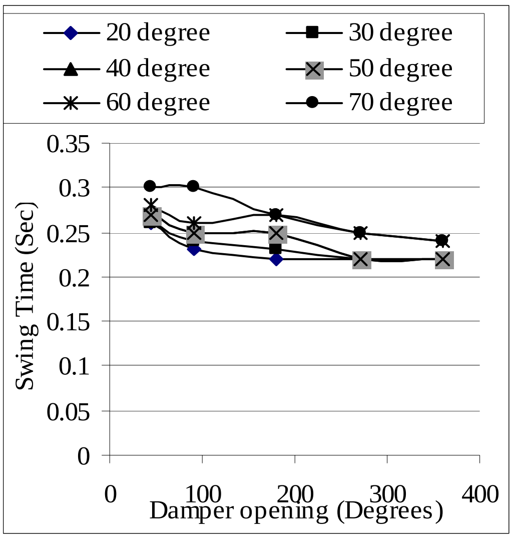

Fig. 7. Graphs of actual time taken to swing from a particular flexion angle. Some variation is being observed for damper opening up to 200 degree. After that the variation is very less

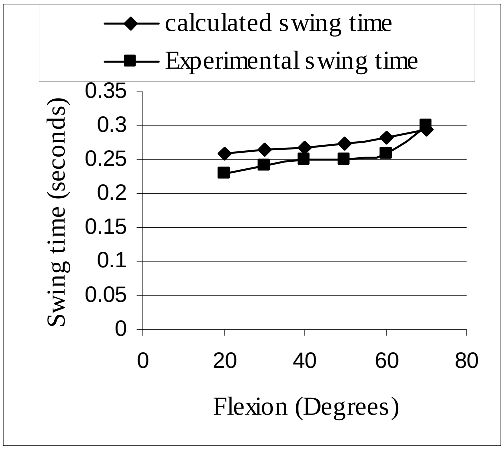

Fig. 8. Comparison between mathematical and actual swing time for 90 degrees damper opening

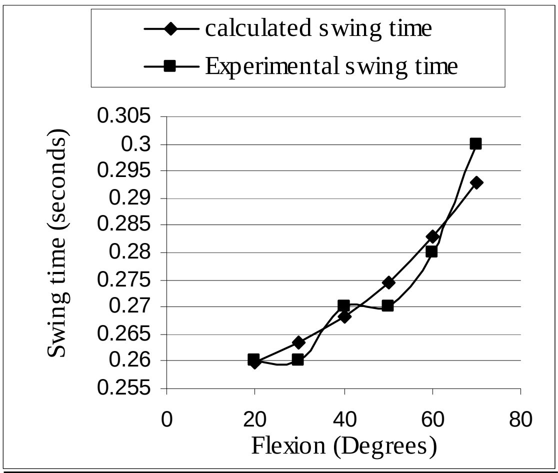

Fig. 9. Comparison between mathematical and actual swing time for 45 degrees damper opening. The mathematical model approaches the actual model with minimal error

Key Findings

The operating range of the selected damper was from 45 degrees opening to 360 degrees only.

Kinematics model can be used to design an AK prosthesis for swing time.

Discussion & Conclusion

Mathematical modelling is a prerequisite for designing and further fabricating any prototype. A simple pivot type knee mechanism is like a pendulum, which is restricted to swing past its neutral position.

Pendulums are devices that execute simple harmonic motion and time period of oscillations can be varied by suitably changing their physical dimensions. In a pendulum motion it is assumed that the friction is at support and the air resistance to motion is negligibly small. In general the various forces involved are: (i) gravitational force, (ii) elastic force, (iii) spring force. Above forces are conservative forces therefore the law of conservation of energy can be applied. The equations of motion, however, can be obtained using the relation 'F = ma'.

The prosthesis can be compared to a compound pendulum instead of a simple pendulum and when a spring and damper system is attached to it, it becomes a case of a second order differential equation, refer - equation 2. An acceptable design is one in which the artificial leg swings within controllable limits to match swing time of able leg.

A compound pendulum though can be compared to a simple pendulum whose time period is given by relation

T= 2 * pi (le/g) ^ 0.5,

where le is the equivalent length and is expressed as: le = ko^2/b; here *ko* is the radius of gyration with respect to the axis of rotation through the pivot, and b is the distance of the centre of gravity from the pivot joint. But this relation does not give correct results when a damper – spring system is added to it.

The spring being nonlinear, experimental results can best be used for a prototype design. Therefore, experimental data has been used to do mathematical calculations using kinematics model approach.

It is concluded that the kinematics model proposed, approaches the actual prototype swing motion. The graphs and results presented in this paper can also be used for selecting motor speeds for active knees. For active knees with motor controls, the controller needs the value of time in which swing is to be completed. The data available in the graphs can be used for this purpose and further motor speed selection.

Declaration of interest

The authors report no conflicts of interest. The authors alone are responsible for the content and writing of the paper. The hardware developed which has been used for doing the testing has been filed for patent in India, under patent Application No.: 886/DEL/2009, Date of Application: 29-04-2009. A National Award has also been given by Ministry of Social Justice and Empowerment, Government of India in public recognition for development of new cost-effective product for manufacture aimed at improving the life of persons with disabilities on 03-December 2009.

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