Effect of Shoe Weight on Swing of Prosthetic Components for Trans-femoral Amputees

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Abstract

The purpose of this study is to determine the effect of different weights of shoes, which amputees may wear on the swing time of artificial leg. A CAD model was analysed and simulated to find swing time during extension phase for 10 – 70 degrees swing angle. A prototype of a pneumatic damper controlled above knee prosthesis with optical encoder based angle sensors fitted on knee joint, was developed in-house and was interfaced with an Atmel S52 micro-controller based data acquisition system. The prosthesis was fitted on a specially designed stand and was made to swing from a specific flexion angle to normal straight position for different weights of shoes. The results showed that as weight of shoe increased swing time increased. It was concluded that weight of shoes to be worn by amputees will affect prosthesis behaviour and reason for such prosthesis behaviour may be communicated to the users.

Key words - Shoe Weight, AK Prosthesis, Swing Time, Dynamic Analysis.

Introduction

People with above knee amputations and using mechanical or automated swing control prosthesis, will feel away from normal walking as the weight of shoe worn by them changes due to change in swing time of the artificial limb. The change in swing time has been noticed and has been validated through CAD modelling on ProE software. The data can be used by fully automatic controlled prosthesis, which will be able to adjust damper values accordingly. Bateni *et al*., (2004), conducted a study to determine the effect of different weights of modular prosthetic components (steel and titanium) on the gait of transtibial amputees. They tested gait parameters of five male unilateral transtibial amputees between the ages of 32 and 77 years in a laboratory setting using both steel and titanium components and studied the kinetic and kinematics gait parameters using a four-segment link segment inverse dynamic model. They also determined physiological cost index (PCI) from measures of walking speed and heart rate using a monitor and found that the PCI was greater when using steel components than when using titanium for all participants ($p = .002$). Speed of walking during PCI tests tended to be greater when the steel components were replaced with titanium (from 1.06 to 1.18 m/s, $p = .026$). Hale (1990), performed a study to determine the effect of varying prosthetic shank mass, while maintaining the mass centre location and moment of inertia, on the swing phase kinematics, kinetics and hip muscular effort of free speed above-knee (AK) amputee gait. Despite increases in shank mass from 1.33 to 3.37 kg the AK amputee was able to maintain a consistent swing time and walking speed. As load increased, there were significant changes in the maximum knee and hip displacements. The prosthetic knee Resultant Joint Moment (RJM) was negligible while the shank was accelerating, but was a major contributor during shank deceleration. Van De Veen *et al*., (1987), investigated the dynamic behaviour of an above-knee (AK) prosthesis in the swing phase using a mathematical model and analysed the influence of mass and mass distribution on the maximal stump load and the required energy. The model consisted of a bond-graph model of the prosthesis and a "walking" model, which predicted the walking velocity, step length and the femoral trajectory. Through computer simulation, stick-diagrams of the swing phase and graphs of the variation with time of the hip and stump forces were obtained. They concluded that, because of their comparable weights the influence of the shoe is almost equal to the influence of the prosthetic foot. Thus lightweight shoes should be used with lightweight prosthetic feet in order to add to their advantages. In our previous study (Soni *et al*., 2010), a mathematical model was made to find the swing time of AK prosthesis. In this work, we have found effect of shoe weight on swing time using dynamic analysis on Mechanism module of Pro-E software and further experimental validated. For validation, a micro-controller based data acquisition system was designed and was fitted to the prototype. The prosthesis fitted on a specially designed stand was made to swing from different flexion angles and readings of swing time were taken.

The main aim of a doctor providing above knee prosthesis solution to a patient, is to ensure comfortable and near to normal walking experience. The patient may complaint of over weight of foot and shoe, but it may not just be weight, but the flexibility of the foot and thus shoe which also affect gait. Even foot design has an impact on comfort level of the amputee. Ronald *et al*., (1995), suggested that the intact lower extremity is susceptible to excessive floor impact, and that prosthetic foot design can have an effect on the magnitude of the vertical forces experienced by the limb. The sound limb loading with the Flex-Foot was significantly less than the SACH foot. There is also relation between walking speed and swing time and speed control can be best incorporated in fully automatic control above knee prosthesis. There is a notion that the prosthesis weight is only responsible for swing time, but apart from weight of the prosthesis, the swing time is also dependent upon the speed of walking and load carried. Hale (1991), performed a study to determine the effect of walking speed on stride length, stride and swing times, the knee and hip displacement and torque patterns, and the roles of the gravitational forces and interactive moments associated with thigh motion. As per Hale, as walking speed increased, the stride length increased while stride and swing times decreased. The maximum knee flexion and initial hip flexion peaks increased as walking speed increased. As walking speed increased, the knee flexor torque during terminal swing increased.

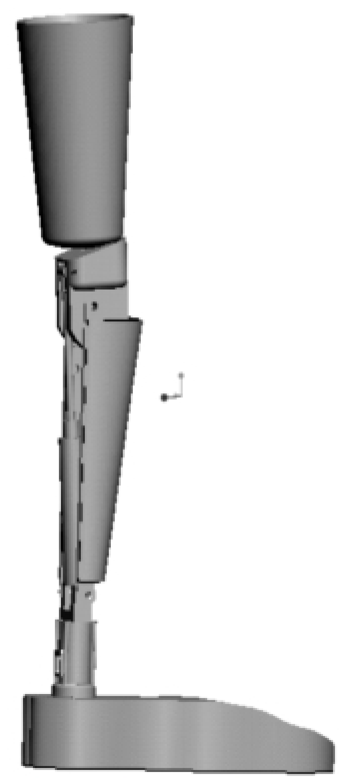

Figure 1: A Pro-E CAD model of the prosthesis used for Dynamic Analysis.

Research into bipedal walking robots has also provided contribution into design of above knee prosthesis solutions for rehabilitation of the impaired. Wisse *et al*., (2006), in their research into bipedal walking machines, stated that the key engineering problem is to keep the weight of the actuation system small enough. For their 2D prototype robot named MIKE, the problem was solved by applying pneumatic McKibben actuators on a passive dynamic biped design. The result was a fully autonomous biped that could walk on a level floor with the same energy efficiency as a human being.

CAD modelling has become the design technique of the day. A dynamic analysis approach offered by CAD soft wares available today are reliable and are being used by engineers extensively to simulate the results prior to development and testing in real world. CAD has been extensively used for structural analysis and socket design, but mechanism design applications are still unexplored. Our work gives an insight into this application. Bae *et al*., (2007), studied to quantitatively evaluate amputee gait by dynamic analysis of the musculo-skeletal system during level walking and stair climbing. Jaroslav Mackerle (1992), listed complete bibliography of conference proceedings papers and theses / dissertations on the finite element (FE) and boundary element (BE) applications in different fields of biomechanics between 1976 and 1991. Wang *et al*., (1992), stated that conventional designs of an above-knee prosthesis are based on mechanisms with mechanical properties (such as friction, spring and damping coefficients) that remain constant during changing cadence and since the non-linear and time-varying dynamic coupling between the thigh and the prosthetic limb is high during swing phase, an adaptive control is necessary to employ to control the knee joint motion and thus an active knee is a necessity.

On the other hand active knee joints are not popular due to need of a power source all the time. Passive knees are sold the most and thus there is need to concentrate research on passive knees also. Gregorio *et al*., (2004), modelled ankle passive motion and stated that the use of (equivalent) planar and spatial mechanisms for the kinematic modelling of joint passive motion can also be successfully utilized for the knee joint. In order to control the swing phase of a passive type above knee prosthesis, weight of the prosthesis plays an important role. Tsai *et al*., (1986), developed a detailed dynamic model of the stump-prosthesis system for an above knee amputee to examine the influence of controls and design parameters on the limb system performance during the swing phase of gait. Their simulations suggest that

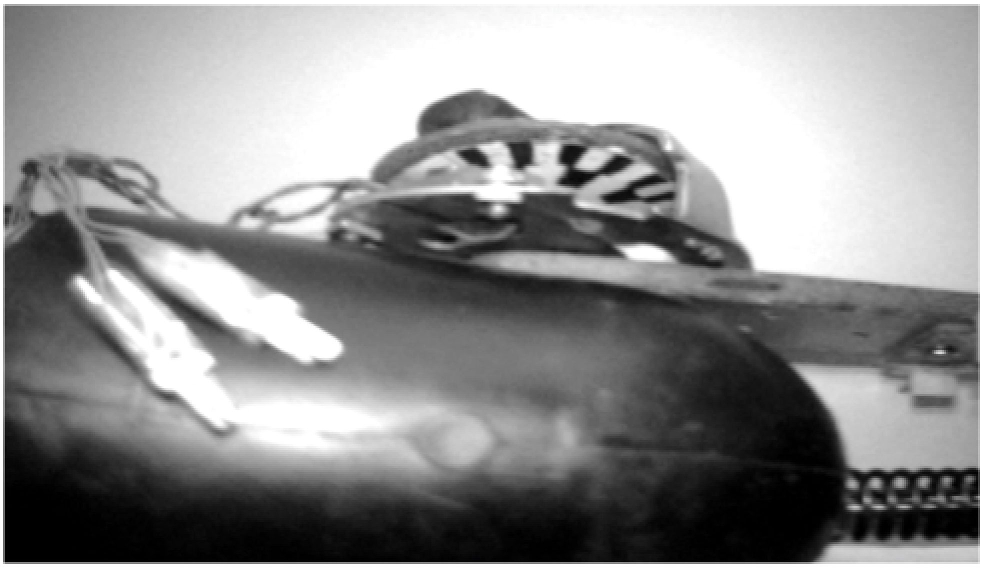

Figure 2: A photograph showing optical encoder fitted to an above knee prosthesis used for angle sensing.

lightweight prosthesis designs do not perform as well as heavier designs. On the contrary, since light - weight will consume less body energy and will be more comfortable and with CAD modelling technique, which is available today, this problem is solvable and an optimum design can be easily developed. The prosthesis developed by us and experimented upon is a passive type with pneumatic damper control system and does not weight more than 2.5 kgs and swings well to match normal cadence.

Rafael *et al*., (2008), presented a review of current researches for development of knee prosthesis. Hao *et al*., (1969), did a systematic analysis to simulate a mathematical model for the control mechanism of an above-knee leg prosthesis and used the CSMP-360 continuous system-modelling program to study various parameters, from linear to piecewise-linear controls. Carli *et al*., (1996), gave an approach to the kinematic analysis of knee joint prosthesis. Frank *et al*., (2006), presented a paper describing the design of an above-knee prosthesis with actively powered knee and ankle joints, both of which are actuated via pneumatic actuators. Our prosthesis is a passive knee joint with single axis motion in sagital plane.

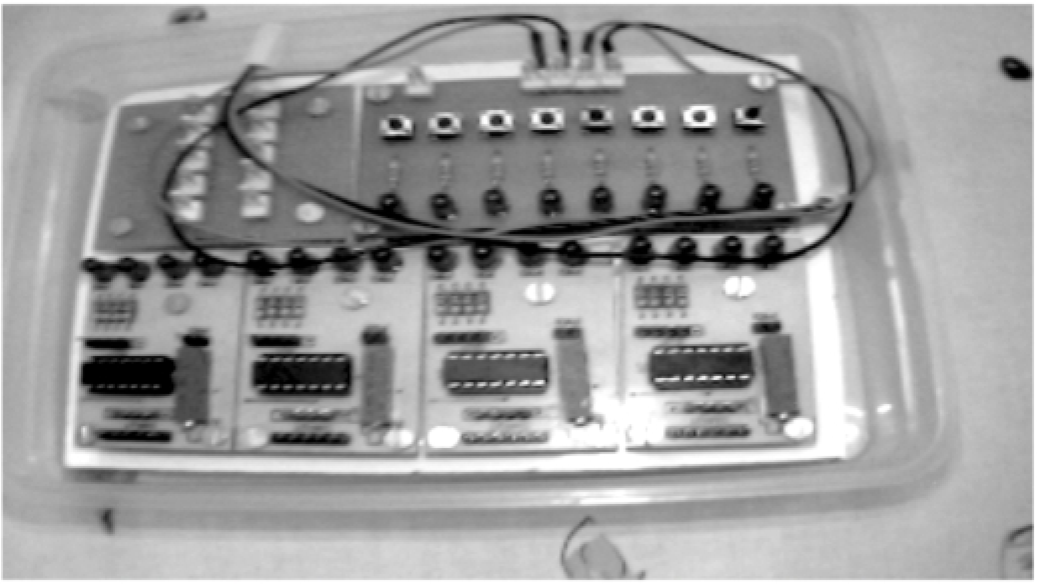

Figure 3: A photograph of operational amplifiers circuit used for interfacing optical encoder to controller.

Knowledge gathered after literature available from sources mentioned above, helped to design it. CAD simulation & experiments were conducted to find effect of shoes on swing time. The results have shown minimal errors between simulated and experimental results and have also generated data of effect of shoe weight on swing time.

Methods

Dynamic analysis offered by CAD soft wares is a simulation tool and one offered by Pro Engineer software was used to design this mechanism. For simulation, the knee and socket were made as ground and a spring and damper assembly along with foot was made as the swinging part. The model made for simulation is shown in Figure 1. The complete assembly has all the parts with same specifications as in real model, including shank strips, cover, foot and shoe. The foot design was approximated to include the shape and weight of shoe, used in prototype. The spring constant of the spring and damper co-efficient of the damper used in the prototype were not available, so these specifications had to be experimentally found and this work has been detailed in our previous work (Soni *et al*., 2010). There after, these values were used to do mechanism design study in Pro E software. Spring constant in compression was taken as

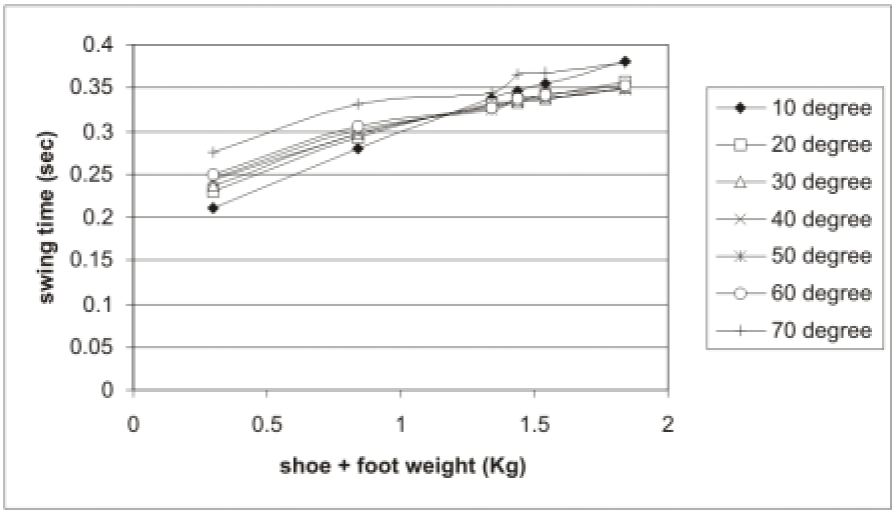

Figure 4: The graph shows simulation results of dynamic analysis done on Pro-E software to find swing time of the artificial leg by increasing shoe weight worn on AK artificial leg.

15 N/mm and damper constant at complete throttle opening was taken to be 0.14 Ns/mm. For the present study, shoes weight was changed every time and dynamic analysis was performed for 10 to 70 degrees flexion angle. Total weight of swinging part of the prototype without shoes was1.6 kg and shoe weight was 540 grams. Extra weights were put on the shoes, so that shoe weight varied from 540 grams to 1540 grams and weight of swinging part thus varied from 1.6 kg to 3.14 kg. Only external gravity load was applied and spring force was inbuilt in the model. The model was thereafter made to swing from the specified flex angles to zero flex angle – the normal straight position.

The experimental prototype consisted of an Alimco - India make prosthetic knee that was further modified to include a pneumatic piston with an external spring. The spring gained energy during flexion, and this energy along with gravitational force was used to extend the prosthesis during swing phase of the

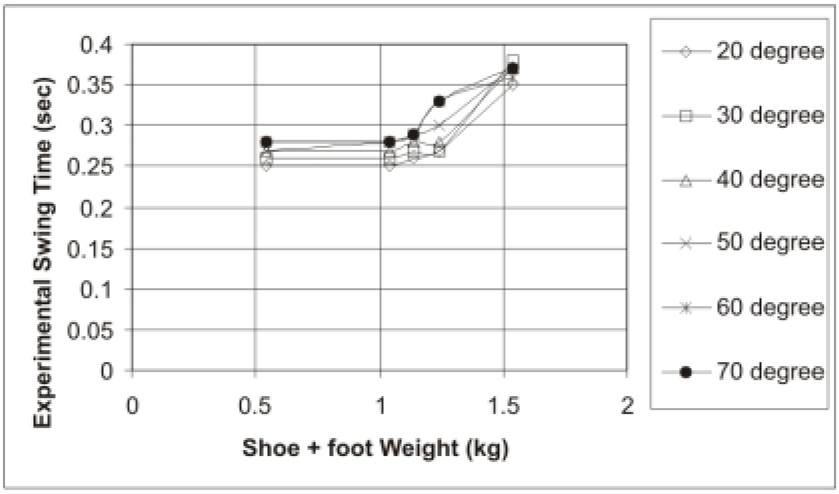

Figure 5: The graph shows experimental results of swinging time of AK prosthesis with increase in weight of artificial foot and shoe.

artificial leg. The control of this energy controls the swing time and a damper was used to control this rate of energy consumption. The complete assembly was there after checked for alignment and was then fixed to a stand with an optical encoder fitted to the knee as shown in figure 2, to find the knee angle during flexion and extension phases. The optical encoder was used as an angle sensor and along with an interface circuitry having operational amplifier as shown in figure 3 was interfaced to an Atmel 8052 micro-controller, which was further programmed to measure and display swing time of the leg.

The swinging part of the leg was displaced mechanically to the specified angle and was released. The software code embedded in controller automatically calculated the swing time and displayed it on an LCD screen.

Results

The Cad Model as shown in figure 1, of the prosthesis was developed for mechanism design and dynamic motion analysis was performed on Pro-Engineer software. The result of dynamic analysis gave swing time of the prosthesis for varying shoe and foot weights and has been detailed in figure 4. As is clear from the figure, as swing angle increases the swing time increases and also as weight of shoe and artificial foot increases, the swing time increases. A prototype, fitted with a micro-controller based data-acquisition system was made to find the swing time experimentally. Figure 5, gives the experimental results of swing time of the prosthesis for different foot plus shoe weights. Clearly, as weight of shoe and foot is increasing, the swing time is also increasing. Minimal changes were observed from 0.5 kg to 1.2 kg, but after that change is significant. This trend is not seen in simulation results and it may be because of hardware limitations. The error between simulated results and experimental results is negligible and thus simulation results have been validated.

Discussion and Conclusion

As is clear from the graph in figure 4, as the weight of shoe worn by the amputee increases, the time taken for swing also increases. While designing for normal walking with an above knee prosthesis the mechanical design has to ensure that the prosthesis swinging time is within controllable limits for normal cadence. Considering normal cadence of 35 double steps per minute, time taken per double step is 1.71 seconds. Approximating swing phase time for one leg to be 40 % of single step time comes to 0.34 seconds. This graph is also validating the design for swing action, since swinging is taking place from 0.2 seconds to 0.38 seconds, which is a basic requirement for any above knee prosthesis for normal plain surface walking. For software analysis, only gravity load was taken and no friction component was added, though it will also be there in actual condition. The rotary motion is about pivot and swing time is dependent upon the moment acting about the pivot. As the weight of shoe plus foot increases, there is shift in centre of gravity location and moment of Inertia of the leg. The equation controlling the motion is a second order differential equation having moment of Inertia, damper and a spring. As weight increases, centre of gravity shifts away from the pivot and the potential energy gained during flexion phase will have an effect to reduce swing time. But, in this prosthesis, there is also an external spring, which gains energy during flexion phase and releases the same during extension phase and this release in energy is responsible for fast swing time. With increased load, the spring will have to push a higher load. It has been observed that prosthesis swing is more affected by energy gained in spring during flexion phase and less because of potential energy gained in this compound pendulum. There is also a damper in the system, but damper constant is not affected by extra weight, as weight is not affecting airflow in the circuit. The slight variation in experimental results- figure5, as compared to simulated results - figure 4 is clear and is obvious. There are factors such as friction and inbuilt error in sensor and data acquisition system, which could not be considered in dynamic analysis, but still the two graphs are almost co-incident. We finally conclude that as weight of shoe and foot assembly increases, the swing time also increases. Wisse *et al*., (2006), in their research into bipedal walking machines, stated that the key engineering problem is to keep the weight of the actuation system small enough. Our work has put another dimension and we state that if even shoe weight changes, there will be a change in swing time and thus comfort level, especially when walking with varying speeds and this along with its reason should be communicated to the users.

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