International Journal of Natural and Engineering Sciences 4 (1): 27-38, 2010
ISSN: 1307-1149, E-ISSN: 2146-0086, www.nobel.gen.tr

A Computer Simulation of the Effects of Mass and Momentum of Inertia Alternation in Individual Muscle Forces During Swing Phase of Transtibial Amputee Gait

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Received: November 13, 2009
Accepted: January 26, 2010

Abstract

An inverse dynamics computer simulation was carried out to investigate the forces of lower extremity muscles in the swing phase of a transtibial amputee gait. With each muscle as an ideal force generator, the lower extremity was simulated as a two-degrees of freedom linkage with the hip and knee as its joints. Kinematic data of hip and knee joints were recorded by a motion analysis system. Through a static optimization approach, the forces exerted by muscles were determined so that recorded hip and knee joint angles were produced. Simulation results showed that when the mass of prosthetic foot is increased, muscle forces increase, too. Also, if the moment of inertia of prosthetic foot is increased, muscle forces will increase, too. However, since prosthetic foot moment of inertia is smaller than that of thigh and shank, its alternation does not have noticeable effect on muscle forces. These results are in accord to experimental and theoretical studies that reported an increase in leg mass and moment of inertia lead to higher electromyography activity of leg muscles, and energetic of walking.

Keywords: Simulation- Swing Phase of Gait- Transtibial Amputee

INTRODUCTION

To investigate the importance of the role played by muscles in the normal swing phase of gait, a lot of research activities have been carried out. Some of them suggest that the forces exerted by muscles in the swing phase may be neglected. For example, Mochon and McMahon, [1] found a range of initial segment angular velocities that could achieve toe clearance without the action of muscles. Also, Mena et al. [2] found that without including moments applied by muscles, a near- normal swing can be simulated. McGeer [3] analyzed and built two- legged passive dynamic machines with knees that could walk down slight slopes without the activities of muscles.

However, the excitations of some muscles in the swing phase are not zero [4]. Therefore, it is reasonable to expect that muscles affect the motions of the swing leg. Piazza and Delp [5] examined the roles of muscles in determining swing phase knee flexion. Riley and Kerrigan [6] used a torque driven forward dynamic simulation to determine whether the rectus femoris and hamstrings muscles contribute to stiff- legged gait if active during the swing phase of the gait cycle. Jonkers et al. [7] analyzed individual muscle function during single stance and swing phase of gait using muscle driven forward simulation. Lim et al. [8] modeled the knee joint to predict the forces of eight main muscle- tendon actuators crossing the knee joint during the swing cycle. Anderson et al. [9] used a three-dimensional dynamic simulation of walking to determine how kinematic conditions at toe-off and muscle forces following toe-off affect peak knee flexion during the swing phase of gait. Arnold et al. [10] analyzed a series of three-dimensional, muscle driven dynamic simulations to quantify the angular accelerations of the knee induced by muscles and other factors during swing phase. Barret et al. [11] employed a forward dynamic simulation of the swing leg to investigate the role played by swing leg muscles. Neptune et al. [12] used muscle-actuated forward dynamics simulations to identify functional and energetic adaptations in individual muscles in response to walking at faster steady-state speeds. Besier et al. [13] used an EMG-driven musculoskeletal model of the knee to estimate quadriceps forces during walking and running.

During the swing phase of transtibial amputee gait, mass and moment of inertia of the foot affect muscle forces, and walking energetic. Experimental records of leg muscles EMG activity and metabolic rate of walking of healthy subjects [14] and energy cost of walking of transtibial amputees [15] show that increasing shank and foot mass and moment of inertia increases the EMG activity of muscles. Using an inverse dynamics simulation of swing phase, Hale [16] reported that a heavier prosthetic shank increases the muscular effort of the amputee. Also, through a computer simulation Dabiri et al. [17] reported that during swing phase of transfemoral amputee gait, when mass and moment of inertia of the prosthetic shank is increased, forces of transected leg muscles will increase, too. Nevertheless, no literature was found that reports the effect of prosthetic foot mass and moment of inertia on individual muscle forces of a transtibial amputee. Therefore, as its goal, this paper quantifies the effect of increasing prosthetic foot mass and moment of inertia on the force exerted by individual muscles during a transtibial amputee swing phase of gait.

Since the forces exerted by muscles were target variables, an inverse dynamics simulation was carried out. The lower extremity was modeled as a two degrees of freedom linkage with hip and knee as its joints. Through a static optimization approach, each muscle force was determined so that experimental recorded hip and knee joint angles were produced.

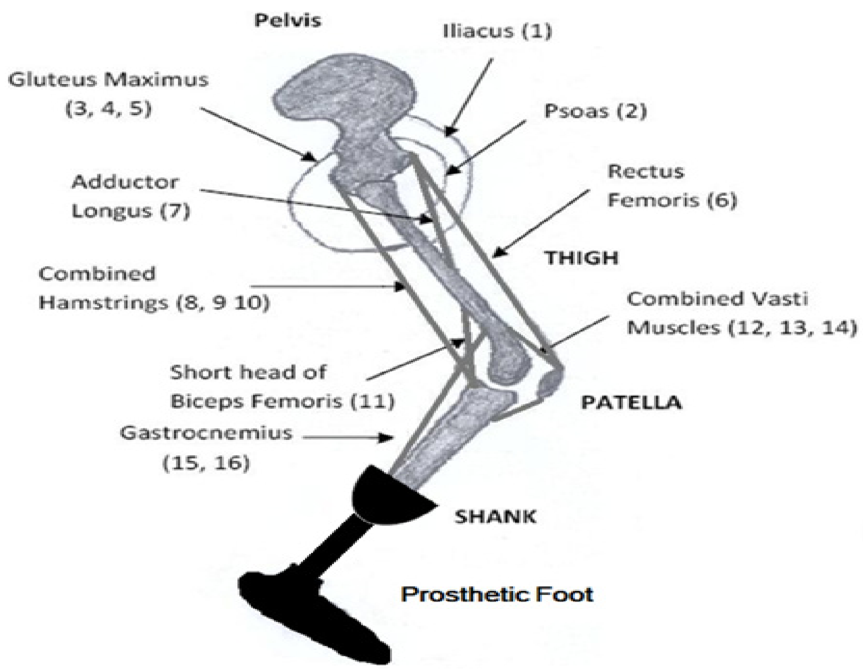
MATERIALS and METHODS

Musculoskeletal Model

The model which is used for lower extremities and their muscles shown in Fig. 1. Only the movements in sagittal plane are considered to be important and it is assumed that there is no rotation between foot and shank. The muscles that are included in the model are: 1- iliacus, 2- psoaa, 3- superior component of gluteus maximus (GMAX1), 4- middle component of gluteus maximus (GMAX2), 5- inferior component of gluteus maximus (GMAX3), 6- rectus femoris (RF), 7- adductor longus (ADDLONG), 8- semimembranosus (SEMIMEM) 9- semitendinosus (SEMITEN), 10- long head of biceps femoris (BIFEMLH), 11- short head of biceps femoris (BIFEMSH), 12- vastus medialis (VASMED), 13- vastus intermedius (VASINT), 14- vastus lateralis (VASLAT), 15- medial head of gastrocnemius, 16- lateral head of gastrocnemius. The origin and insertion point of each muscle is taken from Delp [18]. The mass, geometrical and inertial parameters of the thigh and shank, and foot are presented in Table 1 [5]. To assess the effect of changing foot mass and moment of inertia on muscle forces they were increased (+50%) and decreased (-50%).

To model muscle transaction due to amputation, the distal end of muscles number (15) and (16) were attached to tip of tibia located at the midpoint of shank.

Equations of Motion- The equations of motion were taken from Piazza and Delp [5]:



Dabiri Fig. 1: Schematic of the model.

Table 1. Mass, moment of inertia, and geometrical and inertial properties of the model segments [5].

Parameter	Value
Thigh mass	9.74 kg
Shank mass	3.86 kg
Foot mass	0.99 kg
Thigh moment of inertia	0.167 kg.m^2
Shank moment of inertia	0.060 kg.m^2
Foot moment of inertia	0.005 kg.m^2
Thigh length	0.40 m
Shank length	0.43 m
Thigh distance from proximal end to center of mass	0.20 m
Shank distance from proximal end to center of mass	0.15 m
Foot distance from proximal end to center of mass	0.08 m

$$\begin{bmatrix} \ddot{\theta}_H \\ -\ddot{\theta}_K \end{bmatrix} = M^{-1}C\begin{bmatrix} \dot{\theta}_H^2 \\ \dot{\theta}_K^2 \end{bmatrix} + M^{-1}V\begin{bmatrix} -\dot{\theta}_H\dot{\theta}_K \\ 0.0 \end{bmatrix} + M^{-1}P\begin{bmatrix} \ddot{x} \\ \ddot{y} \end{bmatrix} + M^{-1}G + M^{-1}\begin{bmatrix} M_H \\ -M_K \end{bmatrix} \quad (1)$$

where $\ddot{\theta}_H$ and $\ddot{\theta}_K$ are hip and shank rotational accelerations which are determined from experimental data, $\ddot{x}$ and $\ddot{y}$ are the acceleration of hip joint in horizontal and vertical directions, respectively. M, C, V, P and G G depend upon joint angles and inertial parameters. M_H M_H is the momentum resulted from muscle forces about hip joint, and M_K is the momentum about knee joint.

Static Optimization- Since if one seeks only to estimate muscle forces, the use of dynamic optimization rather than static optimization is not justified [19], the static optimization solution is used. In addition, as taking muscle force- length- velocity properties into account produces results similar to results when they are excluded, each muscle has been treated as an ideal force generator [19]. The performance criterion was chosen as the sum of the squared muscle activations [19]:

$$J = \sum_{m=1}^{MN} (a_m)^2 \quad (2)$$

where J is the performance criterion, MN is the number of muscles, and a_m is the activation of each muscle.

So that muscles can control the motion of hip and knee joints, the equality constraint below is enforced:

$$\begin{bmatrix} \ddot{\theta}_H \\ \ddot{\theta}_K \end{bmatrix} - \begin{bmatrix} \left(\ddot{\theta}_H\right)_{exp.} \\ \left(\ddot{\theta}_K\right)_{exp.} \end{bmatrix} = 0.0 \quad (3)$$

where $\left(\ddot{\theta}_H\right)_{exp.}$, and $\left(\ddot{\theta}_K\right)_{exp.}$

are experimental acceleration of hip and knee joints, respectively.

The experimental accelerations in (3) are computed by twice differentiation of experimental knee and hip joint angles. In addition, the values of muscles activations are bounded between 0 and 1.0.

Experimental Data

Kinematic data of the lower limb during walking were measured by a motion analysis system (WINanalyze 1.4, 3D, Mikromak Gmbh, 1998, Germany). A digital high speed camera (Kodak Motion Corder, SR- 1000, Dynamic Analysis System Pte Ltd, Singapore) was used to record the two-dimensional motion of the body segments taken at 125 frames s^{-1}. Three reflective markers were attached to ankle (lateral malleolus), knee (lateral femoral epicondyle) and hip (greater trochanter).

The values for hip and knee initial velocity and angle for the normal and amputee model used in simulations are presented in Table 2.

Using a backward difference scheme, (1) was solved numerically in MATLAB programming language. Using 100 time steps, on a laptop model Intel® Core™ 2 Tuo CPU T7250 @ 2.00 GHz with 3070 MB RAM, it took about 60 minutes for the healthy model to be run. The execution time for the amputee model was approximately 45 minutes.

RESULTS

To assess the effect of adding mass to individual muscle forces, in Fig. (2) the forces of some muscles when the mass of shank is set to 0.99 (healthy subject foot moment of inertia), 0.495 (50% healthy subject foot moment of inertia) , and 1.98 (200% healthy subject foot moment of inertia) kg are shown. As these figures shows when the mass of shank or its moment of inertia are increased, in order to maintain a near- normal hip and knee angle joints, the amputee should overactivate his muscles.

Also, Fig. (3) shows the forces of muscles when moment of inertia of the shank is set to 0.005 (healthy subject moment of inertia), 0.0025 (50% healthy subject moment of inertia), 0.01 (200% healthy subject moment of inertia) kg. m^2. According to the results shown in this figure, the foot moment of inertia alternation mentioned above does not have noticeable effect on muscle forces.

DISCUSSION

As shown in Fig. (2), when the mass and moment of inertia of foot are increased, the forces of both hip extensors and flexors are increased, too. For a transfemoral amputee, and through experimental records and mathematical modeling, Hale [17] reported that when the mass of shank is increased the muscular effort at the hip joint increases. Through a computer simulation Dabiri et al. [18] reported that during swing phase of transfemoral amputee gait, when the prosthetic shank mass is increased, forces of transected leg muscles will increase, too. Also, Mattes et al. [15] recorded the energy cost of transtibial amputees gait when the mass of the prosthetic shank and foot was varied. They found that as this parameter was increased, the energy cost of walking was increased, too. In addition, recording of metabolic rate of healthy subjects during walking [14], showed when the mass of the leg of a healthy subject increases, his metabolic energy rate increases, too. Consequently, our results in accord with aforementioned literature, show that when the foot mass is increased, the forces of transtibial amputee leg muscles increase, too.

According to Fig. (3), when the moment of inertia of prosthetic foot is changed, the changes in muscle forces are not considerable. Experimental records of muscle activity show that as the moment of inertia of leg is increased, the muscle activities will increase, too ([14] and [15]). Also, mathematical modeling of individual muscle forces showed that as the moment of inertia of the prosthetic shank of a transfemoral amputee leg is increased, muscle activity will increase, too [18]. The reason of the discrepancy between the results shown in Fig. (3) and aforementioned literature is the fact that, as shown in Table (1), the moment of inertia of foot is much smaller than that of the shank and thigh. As a result, the value of moment of inertia of foot has not a noticeable effect on muscle forces. Indeed, close scrutiny of the results shown in Fig. (3) indicate that when prosthetic foot moment of inertia is increased, muscle forces increase, too. Overall, in accord with aforementioned studies, the

Table 2: The hip and knee initial velocity and angle.

Parameter	Value
Normal model hip initial angle	-0.096 (rad)
Normal model hip initial velocity	2.204 (rad s^{-1})
Normal model knee initial angle	0.769 (rad)
Normal model knee initial velocity	4.148 (rad s^{-1})
Amputee model hip initial angle	-0.109 (rad)
Amputee model hip initial velocity	2.080 (rad s^{-1})
Amputee model knee initial angle	0.838 (rad)
Amputee model knee initial velocity	3.600 (rad s^{-1})

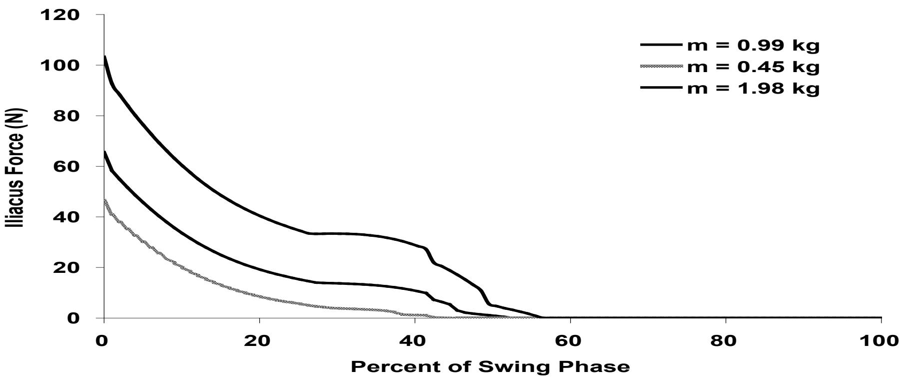


Fig. (2- a): Forces of iliacus muscle for different values of prosthetic foot mass.

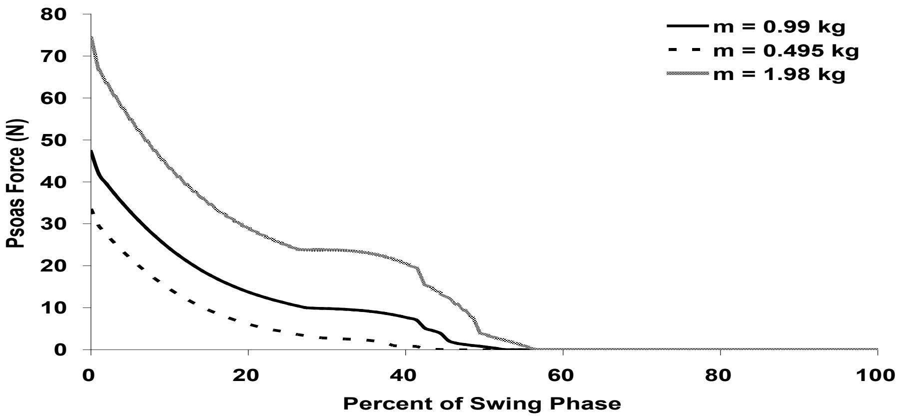


Fig. (2- b): Forces of psoas muscle for different values of prosthetic foot mass.

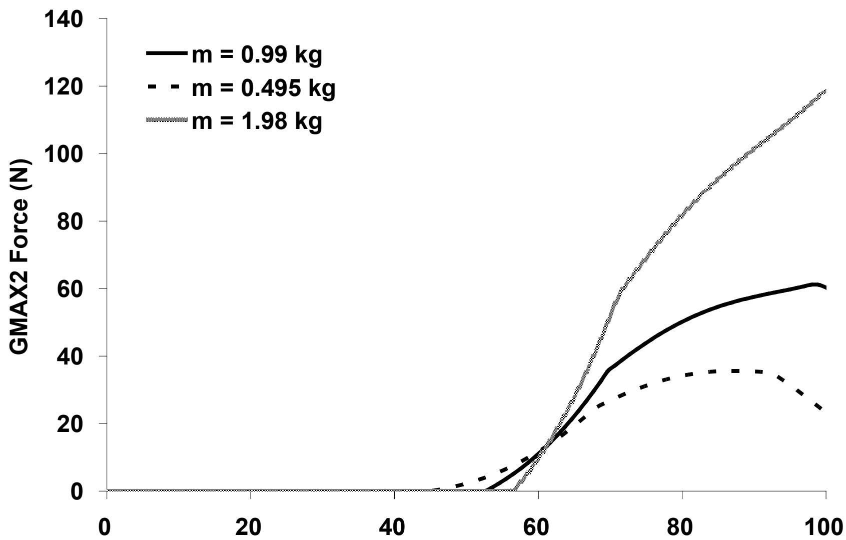


Fig. (2- d): Forces of GMAX2 muscle for different values of prosthetic foot mass.

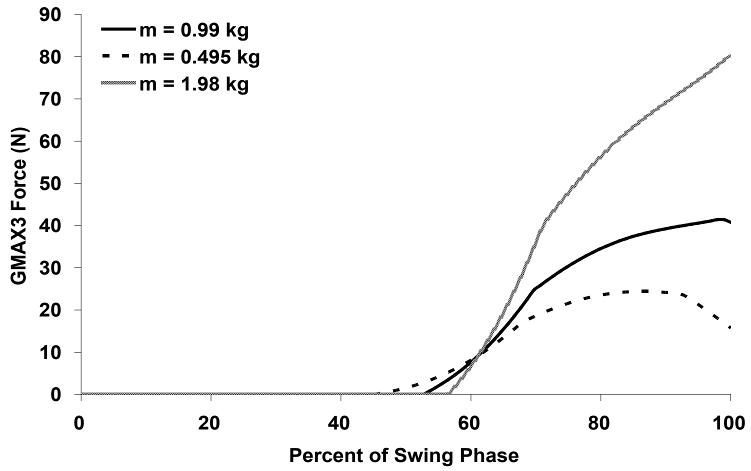


Fig. (2- e): Forces of GMAX3 muscle for different values of prosthetic foot mass.

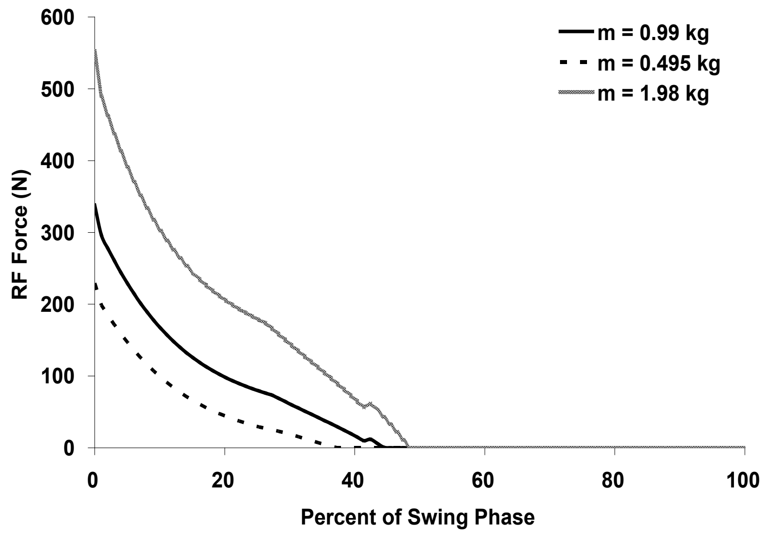


Fig. (2- f): Forces of RF muscle for different values of prosthetic foot mass.

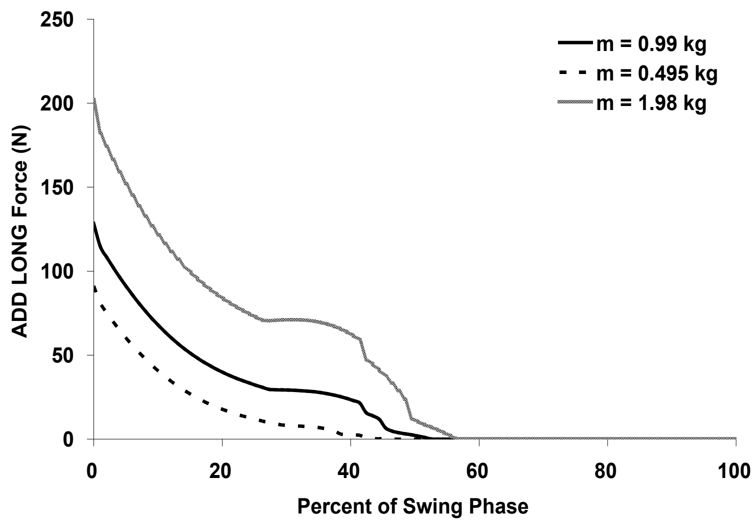


Fig. (2- g): Forces of ADDLONG muscle for different values of prosthetic foot mass.

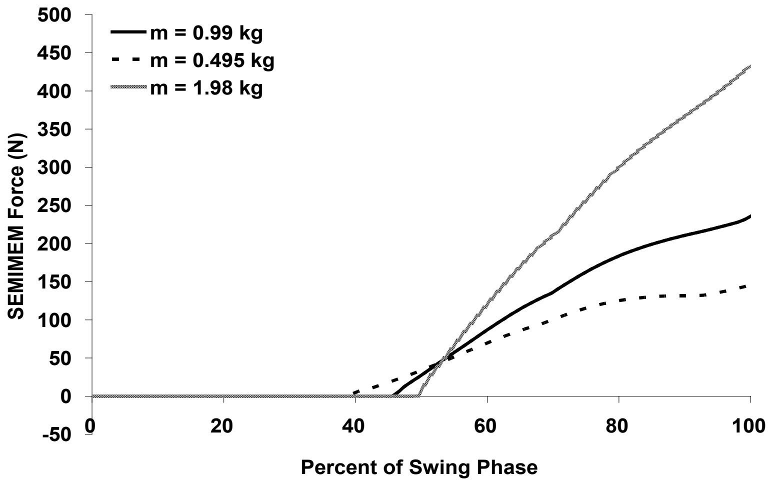


Fig. (2- h): Forces of SEMIMEM muscle for different values of prosthetic foot mass.

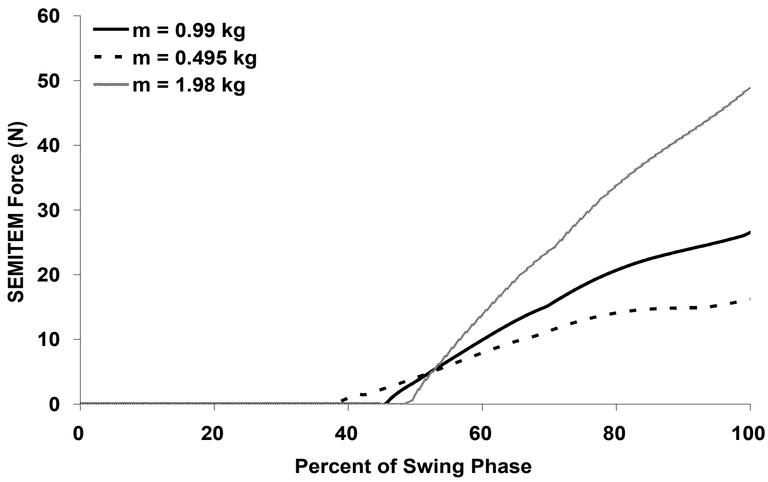


Fig. (2- i): Forces of SEMITEN muscle for different values of prosthetic foot mass.

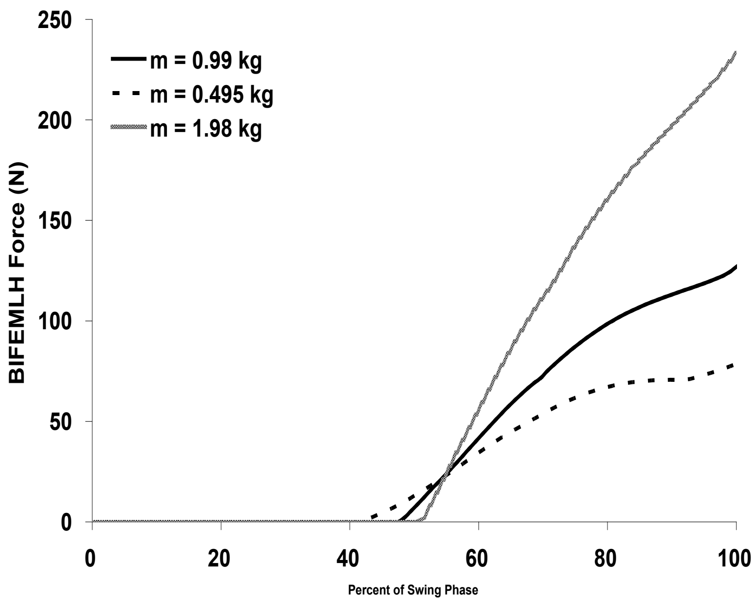
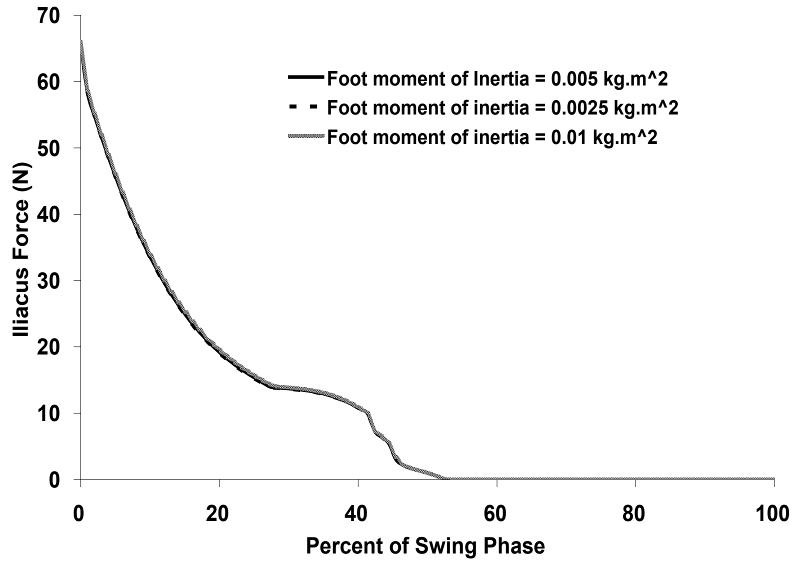
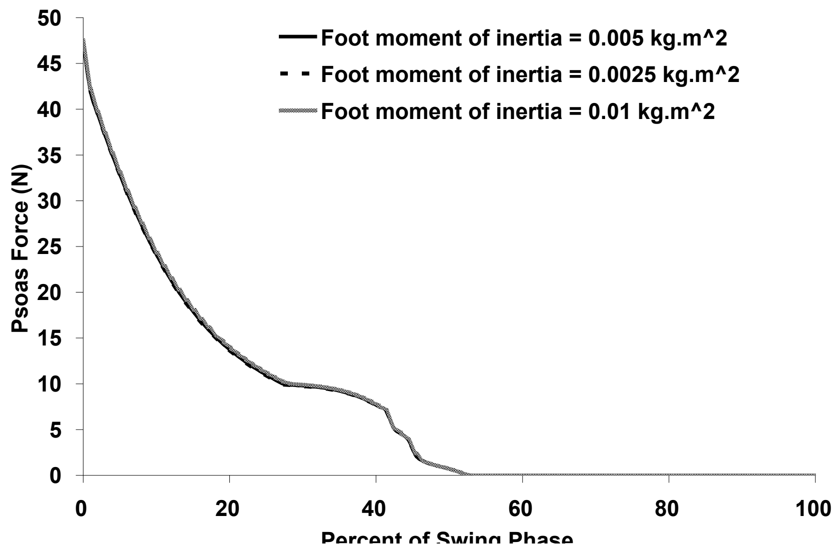


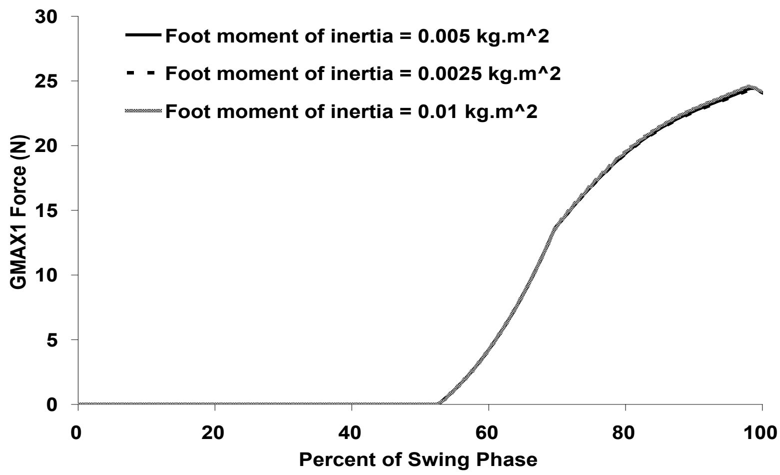
Fig. (2- j): Forces of BIFEMLH muscle for different values of prosthetic foot mass.



Dabiri Fig. (3- a): Forces of iliacus muscle for different values of prosthetic foot moment of inertia.



Dabiri Fig. (3- b): Forces of psoas muscle for different values of prosthetic foot moment of inertia.



Dabiri Fig. (3- c): Forces of GMAX1 muscle for different values of prosthetic foot moment of inertia.

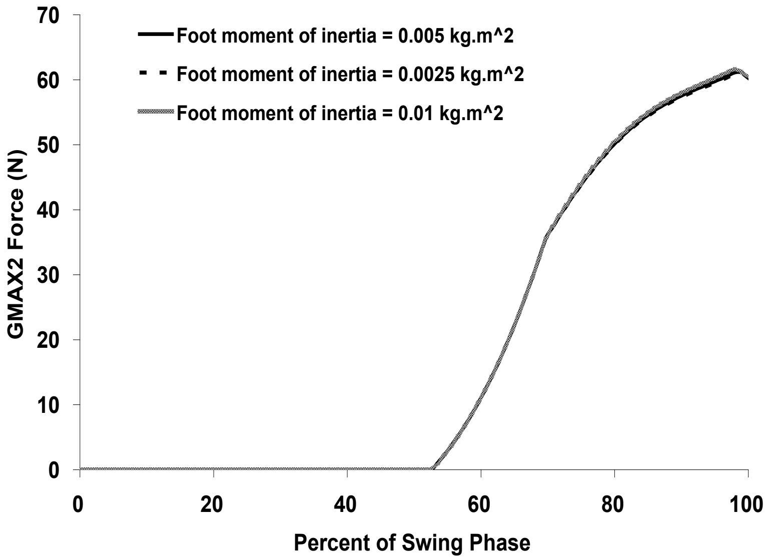


Fig. (3- d): Forces of GMAX2 muscle for different values of prosthetic foot moment of inertia.

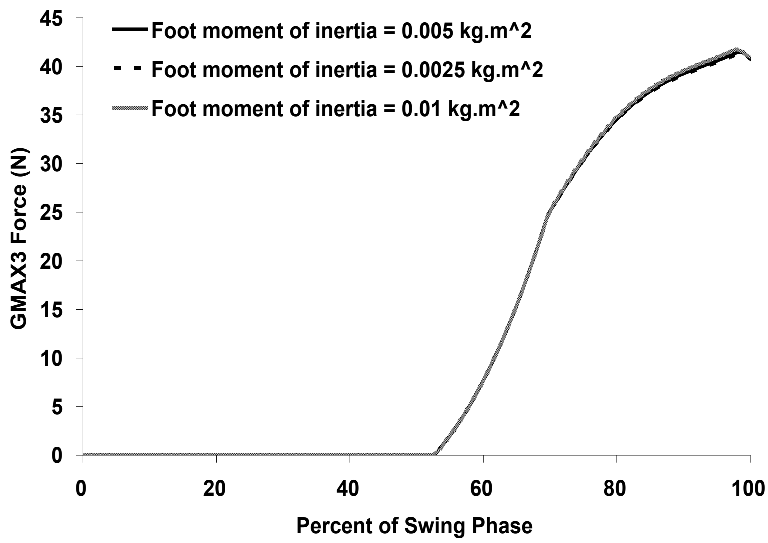


Fig. (3- e): Forces of GMAX3 muscle for different values of prosthetic foot moment of inertia.

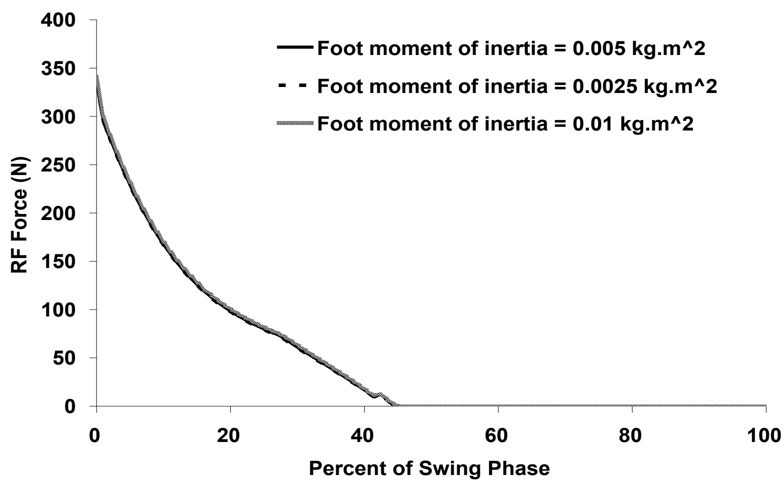


Fig. (3- f): Forces of RF muscle for different values of prosthetic foot moment of inertia.

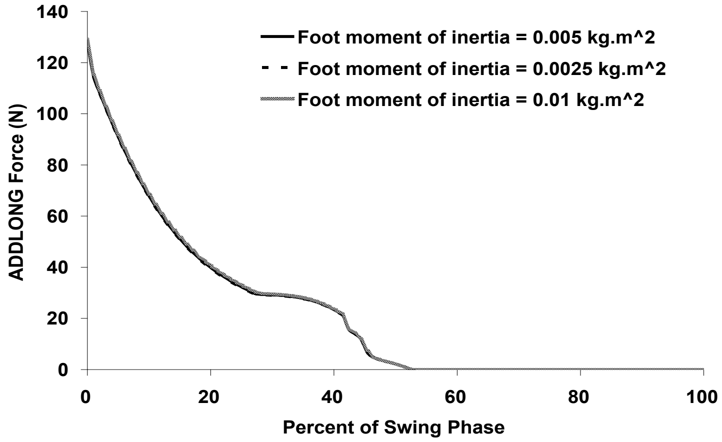


Fig. (3- g): Forces of ADDLONG muscle for different values of prosthetic foot moment of inertia.

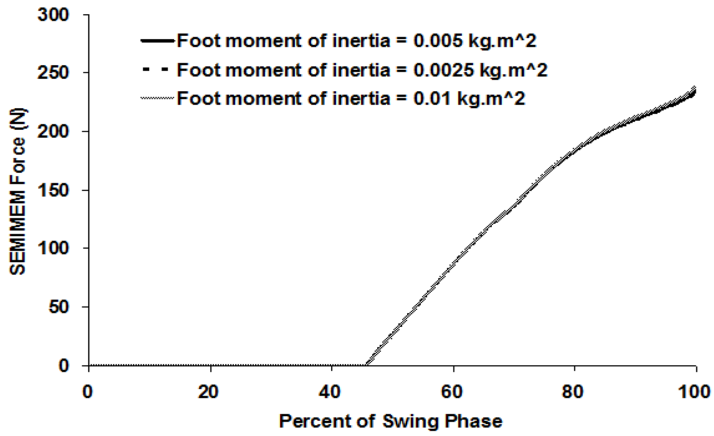


Fig. (3- h): Forces of SEMIMEM muscle for different values of prosthetic foot moment of inertia.

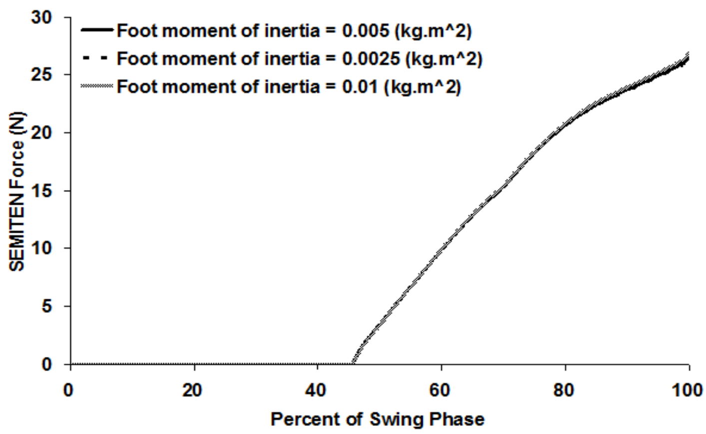


Fig. (3- i): Forces of SEMITEN muscle for different values of prosthetic foot moment of inertia.

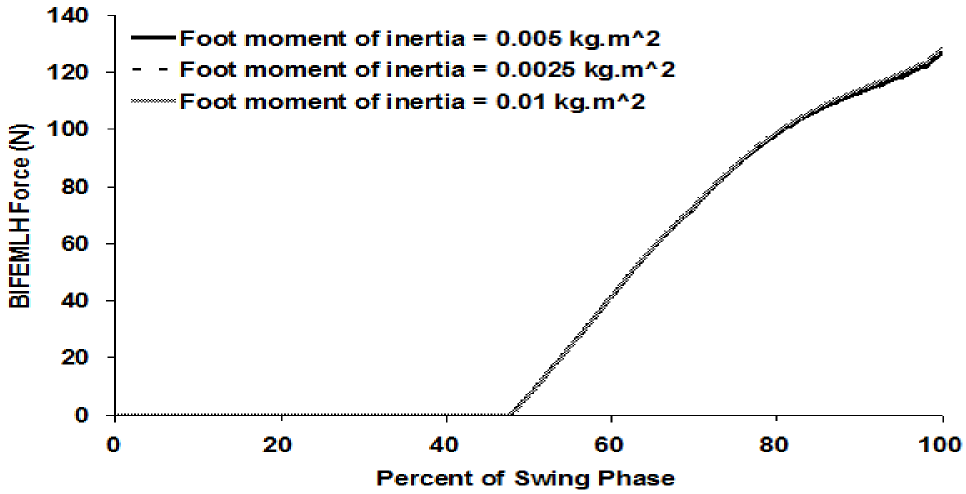


Fig. (3- j): Forces of BIFEMLH muscle for different values of prosthetic foot moment of inertia.

results reported in Fig. (3) show that as moment of inertia of prosthetic foot increases, muscle forces increase, too.

CONCLUSION

A computer simulation of muscle contributions in swing phase of transtibial amputees was presented. According to the results, when the mass or moment of inertia of the prosthetic foot is increased, the individual muscular forces will increase, too. However, since the moment of inertia of prosthetic foot is small in comparison to thigh and shank moment of inertia, its alternation has minor effects on muscle forces. These results are in line with experimental and theoretical studies that reported an increase in leg mass and moment of inertia lead to higher electromyography activity of its muscles, and energetic of walking

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