

Effects of Mass and Momentum of Inertia Alternation on Individual Muscle Forces During Swing Phase of Transtibial Amputee Gait

YAGHOUB DABIRI¹, SIAMAK NAJARIAN^{2*}, M. REZA ESLAMI³,
SAEED ZAHEDI⁴, DAVID MOSER⁵, ELHAM SHIRZAD⁶,
and MOSTAFA ALLAMI⁷

¹*Faculty of Biomedical Engineering, Amirkabir University of Technology, Tehran, Iran*

²*Full-Professor of Biomedical Engineering, Faculty of Biomedical Engineering, Amirkabir University of Technology, Tehran, Iran.*

³*Full-Professor of Mechanical Engineering, Faculty of Mechanical Engineering, Amirkabir University of Technology*

⁴*OBE, FIMechE, Visiting Professor of Biomedical Engineering at the University of Surrey, Technical Director of Chas A Blatchford & Sons Ltd, UK*

⁵*Visiting Researcher of Biomedical Engineering at the University of Surrey, Biomechatronic Control Engineer at Blatchford Product Limited, UK*

⁶*National Olympic and Paralympics Academy, Tehran, Iran*

⁷*Janbazan Medical and Engineering Research Center Tehran, Iran*

Received 17 February 2010/ Accepted 8 March 2010

Key Words: Simulation- Swing Phase of Gait- Transtibial Amputee

A 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. This result is in accord with experimental and theoretical studies that reported an increase in leg mass lead to higher electromyography activity of muscles, and energetic of walking. However, since prosthetic foot moment of inertia is smaller than that of thigh and prosthetic shank, its alternation does not have noticeable effect on muscle forces.

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

MUSCLE FORCES MASS MOMENT INERTIA ALTERNATIONS

(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 prosthetic 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 healthy and prosthetic limb 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.

METHODS

Only the movements in sagittal plane were considered to be important and it was assumed that there was no rotation between prosthetic foot and prosthetic shank. The muscles included in the model were: 1- iliacus, 2- psoas, 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 was taken from Delp (18). To assess the effect of changing prosthetic foot mass and moment of inertia on muscle forces they were increased (+50%) and decreased (-50%).

To model muscle transection due to amputation, the distal end of muscles number 15 and 16 were attached to tip of residual limb.

The equations of motion were taken from Piazza and Delp (5). 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 was used. In addition, as taking muscle force-

length- velocity properties into account produces results similar to results when they are excluded, each muscle was treated as an ideal force generator (19). The performance criterion was chosen as the sum of the squared muscle activations (19). So that muscles can control the motion of hip and knee joints, an equality constraint, according to which the computational and experimental hip and knee angles were equal, was enforced.

Kinematic data of the lower limb during walking were measured by a motion analysis system (WINalyze 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).

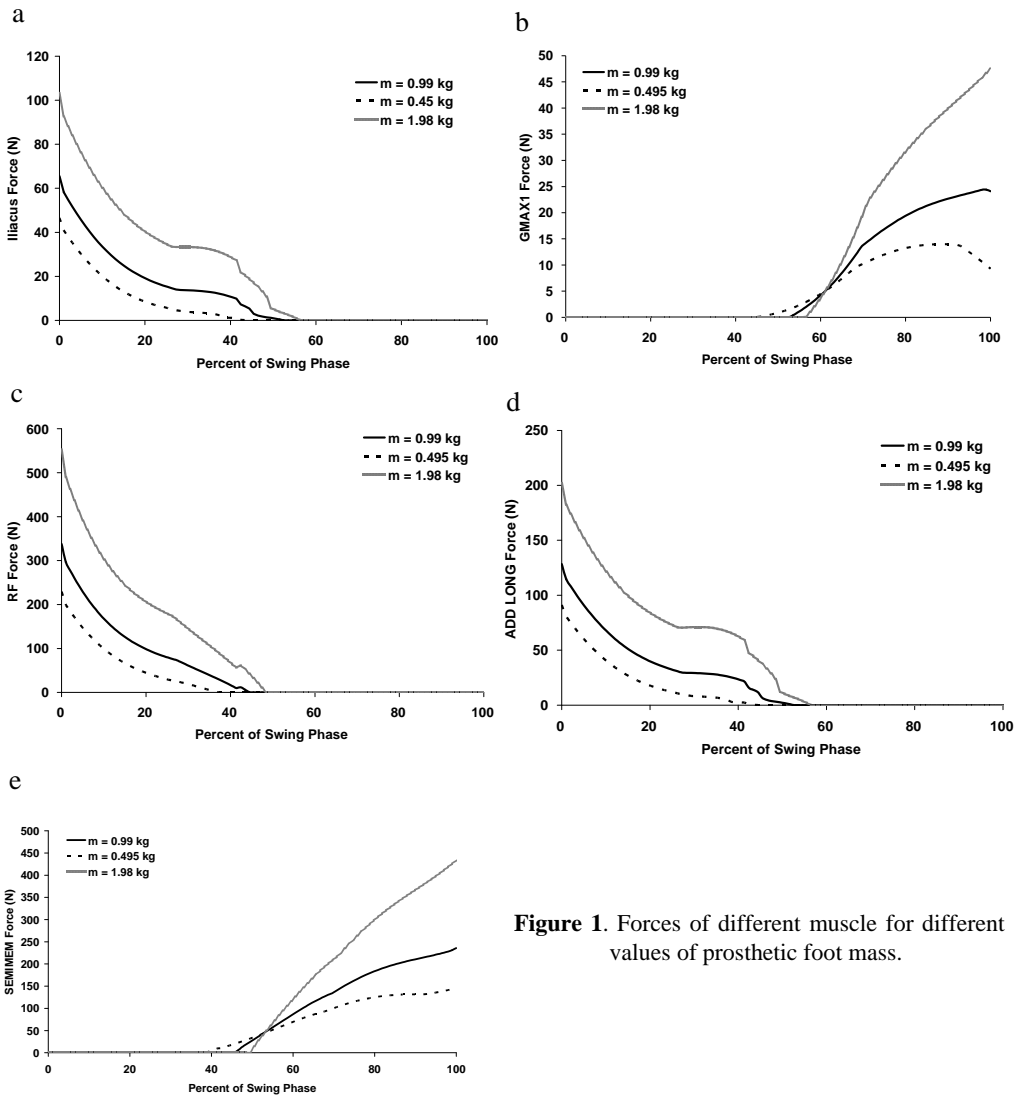


Figure 1. Forces of different muscle for different values of prosthetic foot mass.

MUSCLE FORCES MASS MOMENT INERTIA ALTERNATIONS

RESULTS

To assess the effect of adding mass to individual muscle forces, in Figure (1) the forces of some muscles when the mass of prosthetic foot is set to 0.99 (healthy subject foot mass), 0.495 (50% healthy subject foot mass), and 1.98 (200% healthy subject foot mass) kg are shown. As these figures show when the mass of prosthetic foot is increased, in order to maintain a near- normal angle of hip and knee joints, the amputee should overactivate his muscles.

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

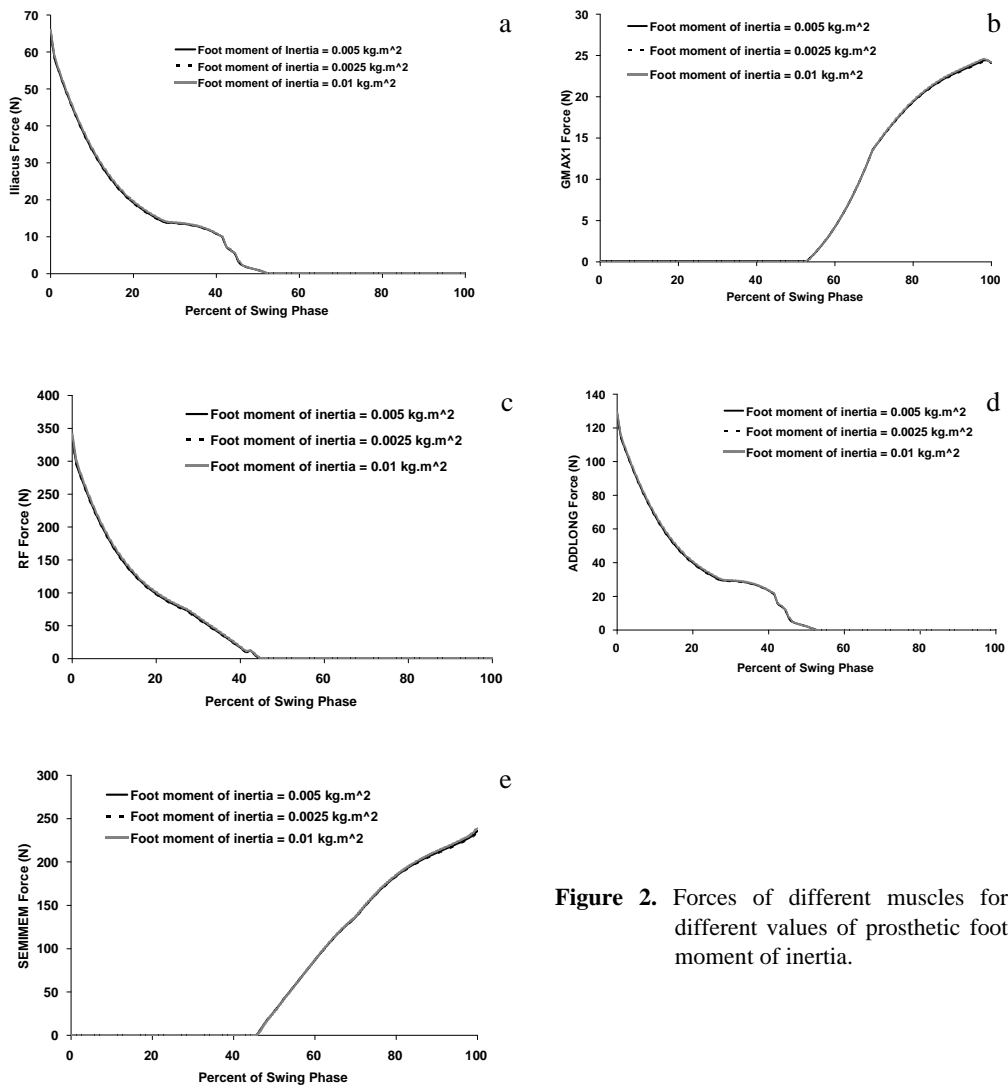


Figure 2. Forces of different muscles for different values of prosthetic foot moment of inertia.

DISCUSSION

As shown in Figure (1), when the mass of prosthetic foot is increased, the forces of both hip extensors and flexors are increased, too. For a transfemoral amputee, and through experimental records and mathematical modeling, Hale (16) reported that when the mass of prosthetic shank is increased the muscular effort at the hip joint increases. Through a computer simulation Dabiri *et al.* (17) 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 limb 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 prosthetic foot mass is increased, the forces of transtibial amputee leg muscles increase, too. According to Figure (2), when the moment of inertia of prosthetic foot is changed, the changes in muscle forces are not considerable. This is because prosthetic foot moment of inertia is smaller than that of thigh and prosthetic shank. As a result, the value of moment of inertia of prosthetic foot has not a noticeable effect on muscle forces.

To improve the results, research is on the way by authors. For example, gait analyses will be conducted to assess the effect of alternation of input variables like hip path, hip angle, and knee angle. Furthermore, using medical imaging techniques such as computed tomography, or magnetic resonance imaging, the attachment points of muscles will be corrected. Also, the effect of interface between residual limb and socket will be included in the model.

REFERENCES

1. **Mochon, S. and McMahon T. A.** 1980. Ballistic walking. *J. Biomech.*, **13**: 49-57.
2. **Mena, D., Mansour J. M., and Simon S. R.** 1981. Analysis and synthesis of human swing leg motion during gait and its clinical applications. *J. Biomech.*, **14**: 823- 832.
3. **McGeer, T.** 1990. Passive dynamic walking. *Int. J. Rob. Res.*, **9**: 62- 82.
4. **Perry, J.** 1992. *Gait Analysis*. 1rd Edition, SLACK, Thorofare, NJ.
5. **Piazza, S. J., and Delp S. L.** 1996. The influence of muscles on knee flexion during the swing phase of gait, *J. Biomech.*, **29**: 723-733.
6. **Riley, P. O., Kerrigan D. C.** 1998. Torque action of two-joint muscles in the swing period of stiff-legged gait: a forward dynamic model analysis, *J. Biomech.*, **31**: 835-840.
7. **Jonkers, I., Stewart C., Spaepen A.** 2003. The study of muscle action during single support and swing phase of gait: clinical relevance of forward simulation techniques, *Gait Posture*, **17**: 97-105.
8. **Lim, C. L., Jones N. B., Spurgeon S.K., Scott J.J.A.** 2003. Modelling of knee joint muscles during the swing phase of gait - a forward dynamics approach using MATLAB/Simulink, *Simu. Mod. Prac. Theo.*, **11**: 91-107.
9. **Anderson, F. C., Goldberg S. R., Pandy M. G., Delp S. L.** 2004. Contributions of muscle forces and toe-off kinematics to peak knee flexion during the swing phase of normal gait: an induced position analysis, *J. Biomech.*, **37**: 731-737.
10. **Arnold, A., Thelen D., Schwartz M., Anderson F., Delp S.** 2007. Muscular coordination of knee motion during the terminal-swing phase of normal gait, *J. Biomech.*, **40**: 3314-3324.
11. **Barret, R. S., Besier Th. F., Llyod D. G.,** 2007. Individual muscle contributions to the

MUSCLE FORCES MASS MOMENT INERTIA ALTERNATIONS

- swing phase of gait: An EMG-based forward dynamics modelling approach, *Simu. Mod. Prac. Theo.*, **15**: 1146-1155.
12. **Neptune, R. R., Sasaki K., Kautz S. A.**, 2008. The effect of walking speed on muscle function and mechanical energetic, *Gait Posture*, **28**: 135-143.
 13. **Besier, Th. F., Fredericson M., Gold G. E., Beaupre´ G. S., Delp S.**, 2009. Knee muscle forces during walking and running in patellofemoral pain patients and pain-free controls, *J. Biomech.*, **42**: 898-905.
 14. **Browning, R. C., Modica J. R., Kram R. , Goswami A. ,** 2007. The effects of adding mass to the legs on the energetics and biomechanics of walking, *Med. Sci. Sports. Exerc.*, **39**:515-525.
 15. **Mattes, S. J., Martin Ph. E., Royer T. D.**, 2000. Walking symmetry and energy cost in persons with unilateral transtibial amputations: Matching prosthetic and intact limb inertial properties, *Arch. Phys. Med. Rehabil.*, **81**: 561-568.
 16. **Hale, S. A.**, 1990. Analysis of the swing phase dynamics and muscular effort of the above- knee amputee for varying prosthetic shank loads, *Prosthet. Orthot. Int.*, **14**: 125-135.
 17. **Dabiri, Y., Najarian S., Zahedi S., Moser D., Shirzad E.**, 2009. Muscle Contributions in the Swing Phase of Transfemoral Amputee Gait: An Inverse Dynamics Approach, *Research Journal of Biological Sciences*, **4**: 1076-1084.
 18. **Delp, S.L.**, 1990. Surgery simulation: a computer graphics system to analyze and design musculoskeletal reconstructions of the lower limb. Dissertation, Stanford University, CA, USA.
 19. **Anderson, F. C. and Pandy M. G.**, 2001. Static and dynamic optimization solutions for gait are practically equivalent, *J. Biomech.*, **34**: 153-161.