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Muscle Contributions in the Swing Phase of Transfemoral Amputee Gait: An Inverse Dynamics Approach

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Abstract: An inverse dynamics computer simulation was carried out to investigate the functionality of lower extremity muscles in the swing phase of a transfemoral amputee gait. With each muscle as an ideal force generator, the lower extremity was simulated as a two-degree 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 in comparison to a healthy model in a transfemoral amputee model, the hip flexors force exertion duration is longer and hip extensors exert smaller force. These results correspond to experimental records of electromyography activity of leg muscles. In addition, results showed that as the mass and moment of inertia of prosthetic shank is increased individual muscle forces will increase. This is in accord with the previous theoretical and experimental studies that showed an increase in both biological and prosthetic shank mass and moment of inertia is a cause to increased muscle efforts and electromyography activity and energy cost of walking.

Key words: Simulation, swing phase of gait, transfemoral amputee, extremity, joints, muscle

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 (1980) found a range of initial segment angular velocities that could achieve toe clearance without the action of muscles. Also, Mena *et al.* (1981) found that without including moments applied by muscles, a near-normal swing can be simulated. McGeer (1990) 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 (Perry, 1992). Therefore, it is reasonable to expect that muscles affect the motions of the swing leg. Piazza and Delpt (1996) examined the roles of muscles in determining swing phase knee flexion. Riley and Kerrigan (1998) 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.* (2003) analyzed individual muscle function during single stance and swing phase of gait using muscle driven forward simulation. Lim *et al.* (2003) 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.* (2004) 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.* (2007) 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. Barret *et al.* (2007) employed a forward dynamic simulation of the swing leg to investigate the role played by swing leg muscles. Neptune *et al.* (2008) used muscle-actuated forward dynamics simulations to identify functional and energetic adaptations in individual muscles in response to

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walking at faster steady-state speeds. Besier *et al*. (2009) used an EMG-driven musculoskeletal model of the knee to estimate quadriceps forces during walking and running.

Dealing with the muscular functionality in transfemoral amputee, this study has two goals. The muscle activity in a transfemoral amputee differs from that in a normal subject. Jaegers *et al*. (1996) reported that the duration and intensity of muscle excitations in a transfemoral amputee are higher than those in a normal subject. Because at present, there is no way to record muscle forces, experimental quantitative assessment of amputation induced alternations in the forces of muscles is not possible. Therefore, as its first goal, this study compares the functionality of muscles of healthy subjects with transfemoral amputees using mathematical modeling.

In addition, the mass and moment of inertia of the prosthetic shank is one of the parameters that affect the muscle functionality. Using an inverse dynamics simulation of swing phase, Hale (1990) reported that a heavier prosthetic shank increases the muscular effort of the amputee. However, Hale (1990) did not quantify the functionality of individual muscles, for he took the role of muscles into account by including their overall moment about hip and knee joints in the equations of motion. In addition, experimental records of leg muscles EMG activity and metabolic rate of walking of healthy subjects (Browning *et al*., 2007) and energy cost of walking of transtibial amputees (Mattes *et al*., 2000) show that increasing shank and foot mass and moment of inertia increases the EMG activity of muscles. Nevertheless, no literature was found that reports the effect of shank mass on individual muscles functionality of a transfemoral amputee. Therefore, as its second goal, this study quantifies the effect of shank mass and moment of inertia on the force exerted by individual muscles in a transfemoral 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-degree 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 in a healthy model is shown in Fig. 1. Only the movements in sagittal plane are considered to be important. Also, it is assumed that hip and knee joints serve as hinge joint and there is no rotation between foot and shank.

The muscles that are included in the healthy model are: iliacus, psoaa, superior component of Gluteus

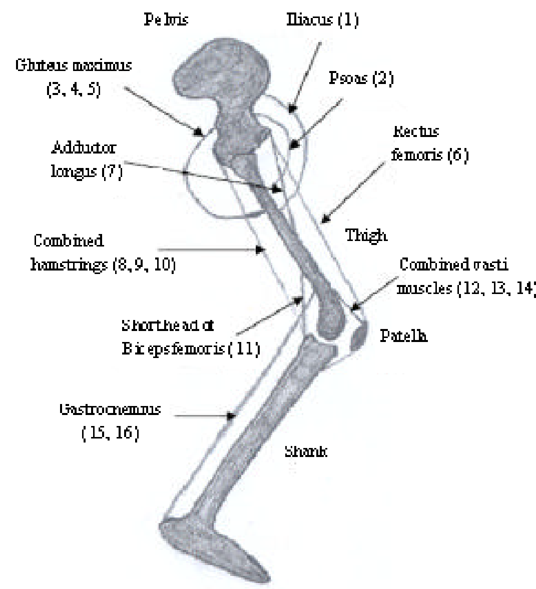

Fig. 1: Schematic of the healthy model

Table 1: The healthy model mass, geometrical and inertial properties (Piazza and Delpt, 1996)

Parameters	Values
Thigh mass	9.74 kg
Shank mass	3.86 kg
Foot mass	0.99 kg
Thigh moment of inertia	0.167 kg m^{-1}
Shank moment of inertia	0.060 kg m^{-1}
Foot moment of inertia	0.005 kg m^{-1}
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

Maximus (GMAX1) middle component of Gluteus Maximus (GMAX2) inferior component of Gluteus Maximus (GMAX3) Rectus Femoris (RF) Adductor Longus (ADDLONG) Semimembranosus (SEMIMEM) Semitendinosus (SEMITEN) long head of Biceps Femoris (BIFEMLH) short head of Biceps Femoris (BIFEMSH) Vastus Medialis (VASMED) Vastus Intermedius (VASINT) Vastus Lateralis (VASLAT) medial head of gastrocnemius, lateral head of gastrocnemius. The origin and insertion point of each muscle is taken from Delp (1990). The mass, geometrical and inertial parameters of the thigh and shank are presented in Table 1 (Piazza and Delpt, 1996).

For the transfemoral amputee model, muscles number 1, 2, 3, 4, 5 and 7 are preserved in the model and muscles number 6, 8, 9 and 10 are preserved partially. Also, it is assumed that there is no rotation between shank and foot and they are modeled as a unit point mass at the center of mass of the shank. The values of shank mass and moment of inertia were set to 2.36 kg and 0.136 kg m^{-2}, respectively (Zarrugh and Radcliffe, 1976).

Equations of motion: The equations of motion are taken from Piazza and Delpt (1996):

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

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 depend upon joint angles and inertial parameters. M_H is the momentum resulted from muscle forces about hip joint and M_K is the momentum about knee joint. For the normal subject this momentum is resulted from muscle forces and for the amputee subject it is resulted from prosthetic knee. In the swing phase of a transfemoral amputee, the prosthetic knee controls the motion in the knee joint. To incorporate the momentum of a prosthetic knee into account a pair of antagonistic muscles is included in the knee joint. In other words, to model the momentum produced by prosthetic knee, a pair of virtual muscles that span knee joint is embedded. This approach is based on the study reported by Hale (1990).

Static optimization: Since if one seeks only to estimate muscle forces, the use of dynamic optimization rather than static optimization is not justified (Anderson and Pandy, 2001) the static optimization solution is used. In addition, as taking muscle force-length-velocity properties into account produces results similar to results obtained when they are excluded, each muscle has been treated as an ideal force generator (Anderson and Pandy, 2001). The performance criterion was chosen as the sum of the squared muscle activation (Anderson and Pandy, 2001):

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

Where:

J = The performance criterion
MN = The number of muscles
a_m = The activation of each muscle

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

$$\begin{bmatrix} \ddot{\theta}_H \\ \ddot{\theta}_K \end{bmatrix} \begin{bmatrix} (\ddot{\theta}_H)_{exp.} \\ (\ddot{\theta}_K)_{exp.} \end{bmatrix} = 0.0 \tag{3}$$

Table 2: The hip and knee initial velocity and angle

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

where, $(\ddot{\theta}_H)_{exp}$ and $(\ddot{\theta}_K)_{exp}$ are experimental acceleration of hip and knee joints, respectively. The experimental accelerations in Eq. 3 are computed by twice differentiation of experimental knee and hip joint angles. In addition, values of muscles activations are bounded between 0 and 1.0.

Experimental data: A male left-side transfemoral amputee volunteered to participate in a motion analysis. He had >12 months experience in using a transfemoral prosthesis with Endolite esprit foot (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) and a Naptesco Hybrid knee (Naptesco Corp., Japan). This prosthetic knee has a microprocessor-controlled pneumatic pressure for swing phase control. The amputee has no other concomitant disabilities and skin complications.

The amputee was asked to walk along a walkway at his natural cadence. Kinematic data of the lower limb during walking were measured by a motion analysis system (WIN analyze 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 sec^{-1}. Three reflective markers were attached to ankle (lateral malleolus) knee (lateral femoral epicondyle) and hip (greater trochanter).

The EMG signal of the muscles was collected by an EMG telemetry system (Mega Electronics Ltd., Kuopio, Finland) at 2000 Hz and processed by an interactive computer program. The values for hip and knee initial velocity and angle for the normal and amputee model were computed from experimental data and are presented in Table 2. Using a backward difference scheme, Eq. 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 min for the healthy model to be run. The execution time for the amputee model was approximately 45 min.

RESULTS AND DISCUSSION

In Fig. 2a-j the forces of muscles are shown for both healthy model and amputee one. As this Fig. 2 shows, there are differences between the forces of these models.

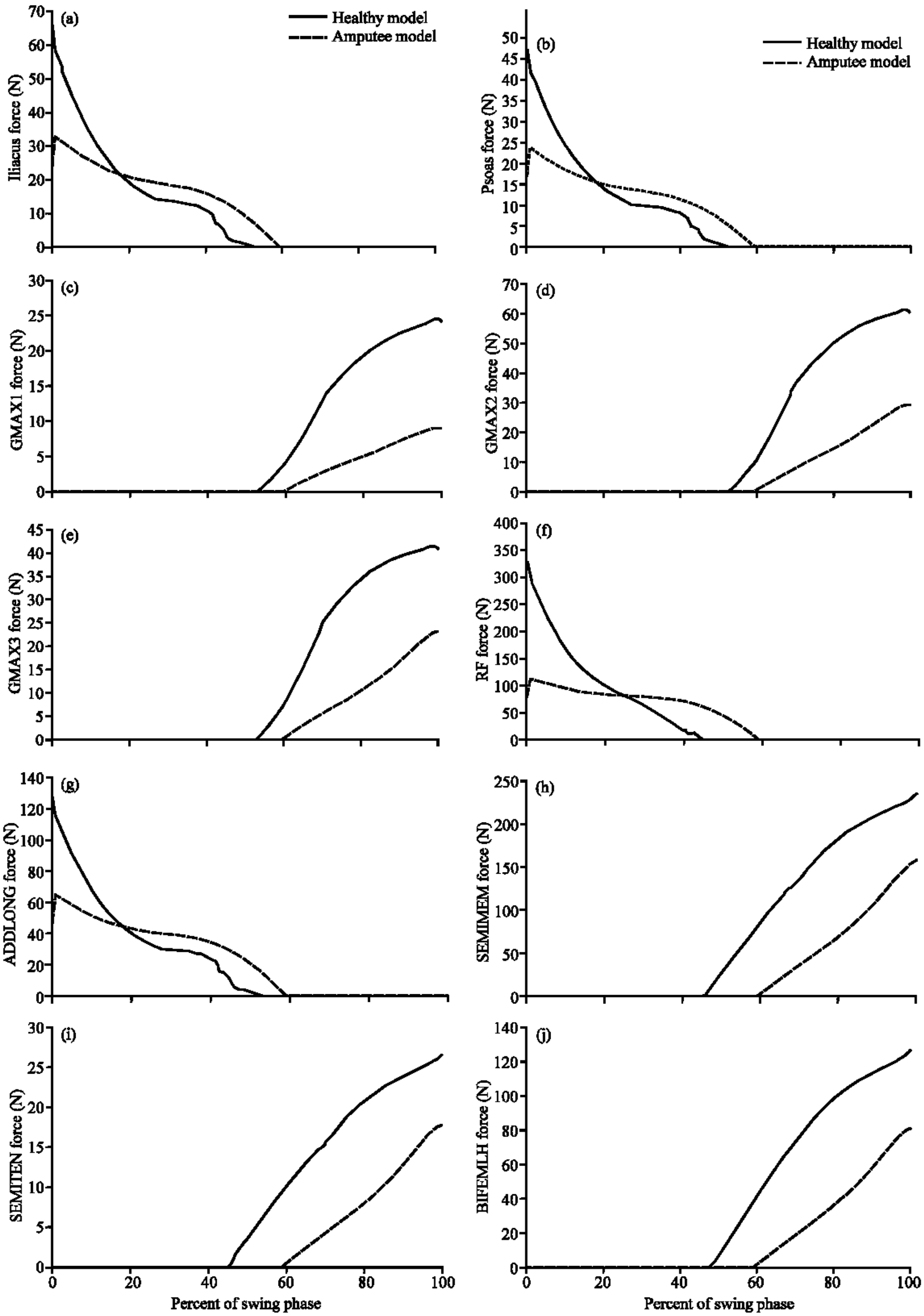

Fig. 2: Comparison between forces of iliacus, psoas, GMAX1, GMAX2, GMAX3, RF, ADDLONG, SEMIMEM, SEMITEN and BIFEMLH muscle in healthy and transfemoral amputee models

In the amputee model, iliacus, psoas, RF and ADDLONG exerts force in a longer duration than healthy model. On the other hand, in the healthy model the duration of force exertion of gluteus maximus and hamstrings is longer in comparison to the amputee one. Also, these muscles exert larger force in comparison to the amputee model.

Since, the model of lower extremity only takes the motions in the sagital plane into account, the muscles can play the role of a hip or knee flexor or extensor. According to Fig. (2) in the amputee model the hip flexor are more active in comparison to the healthy model, but hip extensors are less active. Namely, iliacus, psoas, RF and ADDLONG exert force in a larger duration. But, GMAX and combined hamstrings exert force in a shorter duration, and their mean force is smaller.

According to Jaegers *et al*. (1996) both the intensity and duration of the experimentally recorded EMG activity of RF and ADDLONG in a transfemoral amputee subject are larger in comparison to a healthy subject. However, Jaegers *et al*. (1996) reported these muscles as hip abductor muscles. Considering the results obtained in our simulation, we postulate that the EMG activity of these muscles will have more duration and higher intensity in a transfemoral subject in comparison to a healthy one, for they play a role in both hip abduction and flexion. Since iliacus and psoas are not superficial as RF and ADDLONG no EMG signal has been reported for them. Regarding the results presented in Fig. (2) they act similar to RF in hip flexion and so we suggest that the EMG activity of these muscles will have a longer duration in comparison to a healthy subject.

To validate the simulation results for GMAX force, the EMG activity of this muscle was recorded. The results for both intact and transected leg are shown in (Fig. 3). Since, the EMG activity in a intact limb is more noticeable in comparison to a transected one, considering Fig. 2c-e, the results of simulation seem reasonable.

In Fig. 4 the forces of muscles when the mass of shank is set to 1.6, 2.36, 2.6 and 4.2 kg are shown. Also, Fig. 5 shows the forces of muscles when moment of inertia of the shank is set to 0.06, 0.08, 0.1 and 0.136 $kg\,m^{-2}$. As these Fig. (4-5) show 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.

As Fig. 4 (a-j) and 5 (a-j) show when the mass and moment of inertia of shank are increased, the forces of both hip extensors and flexors are increased, too. Through experimental records and mathematical modeling, Hale (1990) reported that when the mass of shank is increased the muscular effort at the hip joint increases. Also, Mattes *et al*. (2000) recorded the energy cost of transtibial amputees gait when the mass and moment of inertia of the prosthetic shank and foot was varied.

They found that as these parameters were increased, the energy cost of walking was increased, too. In addition, recording of metabolic rate of healthy subjects during walking (Browing *et al*., 2007), showed when the mass of the leg of a healthy subject increases, his metabolic energy rate increases, too. Also, the more distal is the increased load of leg, the greater EMG activity of muscles will be. Therefore, in accord to the results reported by aforementioned literature, the results of our simulation showed that the forces of individual hip flexor and extensor muscles increase when prosthetic shank mass and moment of inertia are increased.

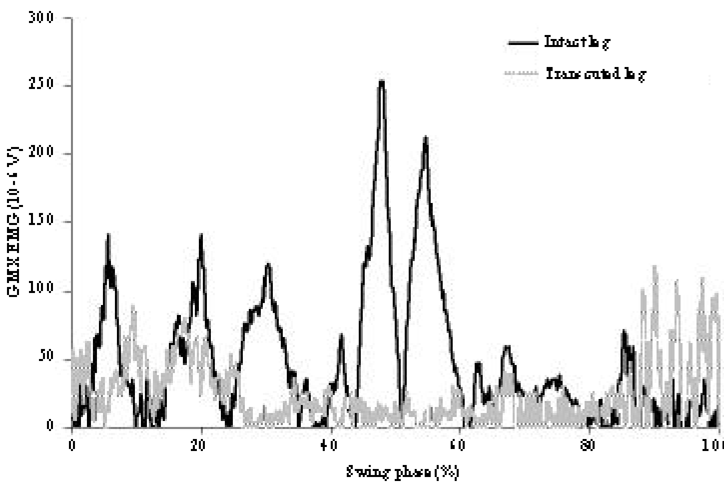

Fig. 3: EMG activity of GMAX in intact and transcuted limb during swing phase of gait

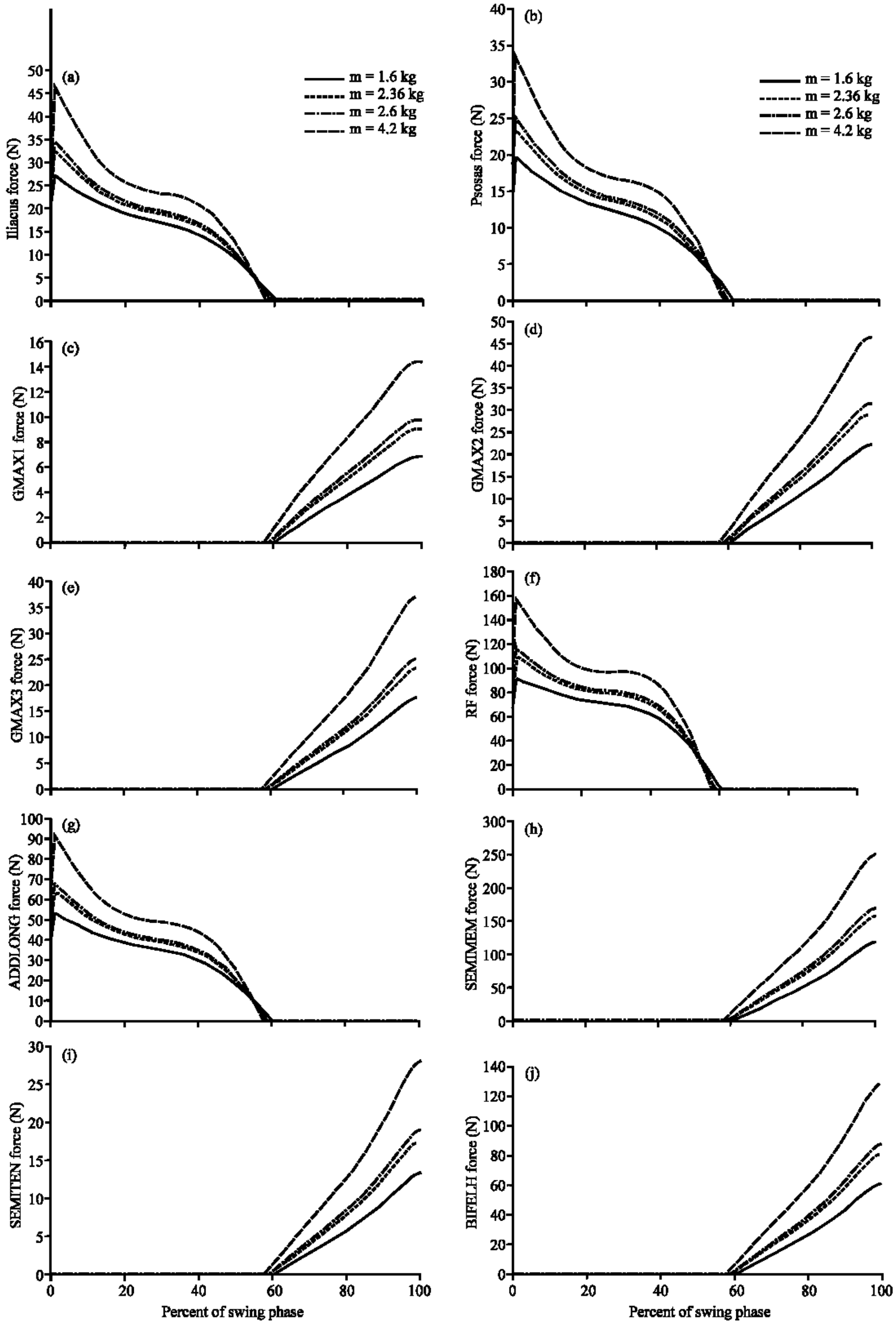

Fig. 4: Forces of iliacus, psoas, GMAX1, GMAX2, GMAX3, RF, ADDLONG, SEMIMEM, SEMITENand BIFEMLH muscle for different values of shank mass

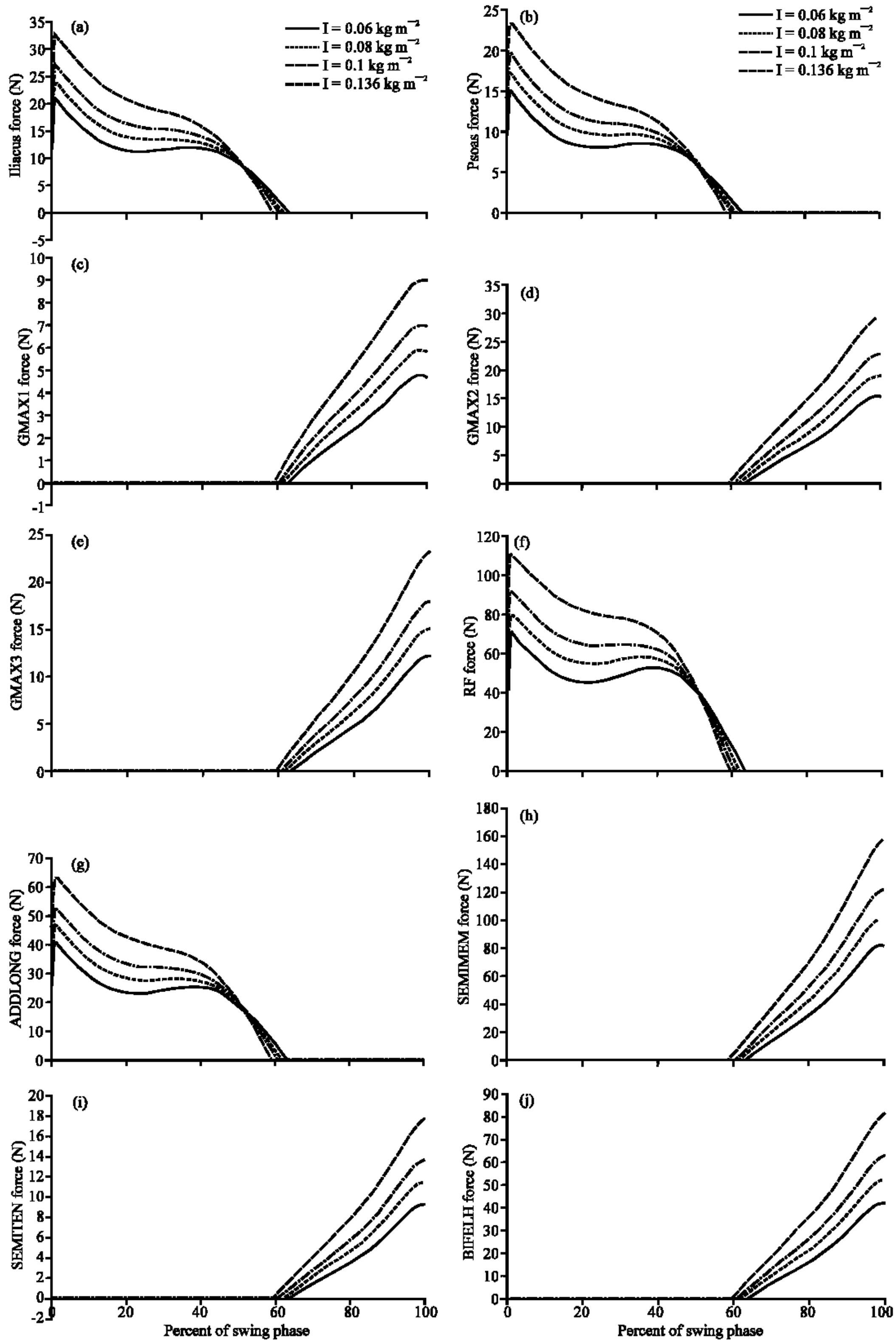

Fig. 5: Forces of iliacus, psoas, GMAX1, GMAX2, GMAX3, RF, ADDLONG, SEMIMEM, SEMITEN and BIFEMLH muscle for different values of shank moment of inertia

CONCLUSION

A computer simulation of muscle contributions in swing phase of transfemoral amputees was presented. According to the results, in comparison to a healthy model the hip flexors of a transfemoral model have longer duration and hip extensors have shorter duration and exert smaller forces. Also, when the mass or moment of inertia of the prosthetic shank is increased, the individual muscular forces will increase, too. These results correspond to previous published literature that report in intact and transected leg, EMG activity of muscles and walking energy rate increase as mass and moment of inertia of the shank is increased.

REFERENCES

Anderson, F.C., S.R. Goldberg, M.G. Pandy and S.L. Delp, 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. Biomechanics, 37: 731-737. DOI: 10.1016/j.jbiomech.2003.09.018. PMID: 15047002. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-4B6KDNB-1&_user=10&_coverDate=05%2F31%2F2004&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=6577bbe47c9803fe7c53881e3b8b0c64.

Anderson, F.C. and M.G. Pandy, 2001. Static and dynamic optimization solutions for gait are practically equivalent. J. Biomechanics, 34: 153-161. DOI: 10.1016/S0021-9290(00)00155-X. PMID: 11165278. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-429XVN8-1&_user=10&_coverDate=02%2F28%2F2001&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=89f4b20c4015c7a51a74a49e2260fcc6.

Arnold, A., D. Thelen, M. Schwartz, F. Anderson and S. Delp, 2007. Muscular coordination of knee motion during the terminal-swing phase of normal gait. J. Biomechanics, 40: 3314-3324. DOI: 10.1016/j.jbiomech.2007.05.006. PMID: 17572431. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-4P0N27G-2&_user=10&_coverDate=12%2F31%2F2007&_alid=1072168448&_rdoc=1&_fmt=high&_orig=search&_cdi=5074&_sort=r&_docanchor=&view=c&_ct=368&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=ad6b92e4add133ff8d1f5a709303a683.

Barret, R.S., T.H.F. Besier and D.G. Lioyd, 2007. Individual muscle contributions to the swing phase of gait: An EMG-based forward dynamics modelling approach. Simulation Modelling Practice and Theory, 15: 1146-1155. DOI: 10.1016/j.simpat.2007.07.005. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6X3C-4P9SNG3-1&_user=10&_coverDate=10%2F31%2F2007&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=74229224b74ff22de5d1b3b0248624e9.

Besier, T.H.F., M. Fredericson, G.E. Gold, G.S. Beaupre and S. Delp, 2009. Knee muscle forces during walking and running in patellofemoral pain patients and pain-free controls. J. Biomechanics, 42 (7): 898-905. DOI: 10.1016/j.jbiomech.2009.01.032. PMID: 19268945. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-4VS9M12-3&_user=10&_coverDate=05%2F11%2F2009&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=d626f64a2fa107e87f13fa7489291c94.

Browning, R.C., J.R. Modica, R. Kram and A. Goswami, 2007. The effects of adding mass to the legs on the energetics and biomechanics of walking. Med. Sci. Sports and Exerc., 39 (3): 515-525. DOI: 10.1249/mss.0b013e31802b3562. PMID: 17473778. http://journals.lww.com/acsm-msse/Abstract/2007/03000/The_Effects_of_Adding_Mass_to_the_Legs_on_the.15.aspx.

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, pp: 89-106. http://nmbl.stanford.edu/publications/index.htm.

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. DOI: 10.3109/03093649009080338. PMID: 2095530. http://www.informaworld.com/smpp/content~db=all~content=a906220309.

Jaegers, S.M., J.H. Arendzen and H.J. de Jongh, 1996. An electromyographic study of the hip muscles of transfemoral amputees in walking. Clin. Orthopaedics and Related Res., 328: 119-128. PMID: 8653944. http://ovidsp.tx.ovid.com/spa/ovidweb.cgi?T=JS&PAGE=fulltext&D=ovft&AN=00003086-199607000-00020&NEWS=N&CSC=Y&CHANNEL=PubMed.

Jonkers, I., C. Stewart and A. Spaepen, 2003. The study of muscle action during single support and swing phase of gait: Clinical relevance of forward simulation techniques. Gait and Posture, 17: 97-105. DOI: 10.1016/S0966-6362(02)00057-7. PMID: 12633768. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T6Y-47G4602-C&_user=10&_coverDate=04%2F30%2F2003&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=4f8e6005820065a03911cb770d16472f.

Lim, C.L., N.B. Jones, S.K. Spurgeon and J.J.A. Scott, 2003. Modelling of knee joint muscles during the swing phase of gait-a forward dynamics approach using MATLAB/Simulink. Simulation Modelling Practice and Theory, 11: 91-107. DOI: 10.1016/S1569-190X(02)00133-8. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6X3C-481MRCH-8&_user=10&_rdoc=1&_fmt=&_orig=search&_sort=d&_docanchor=&view=c&_searchStrId=1072061313&_rerunOrigin=google&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=7a15f022deacbd38437160c4c16db077.

Mattes, S.J., P.H.E. Martin and T.D. Royer, 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. DOI: 10.1016/S0003-9993(00)90035-2. PMID: 10807092. http://www.archives-pmr.org/article/S0003-9993(00)70124-9/abstract.

McGeer, T., 1990. Passive dynamic walking. Int. J. Rob. Res., 9: 62-82. DOI: 10.1177/027836499000900206. http://ijr.sagepub.com/cgi/content/abstract/9/2/62.

Mena, D., J.M. Mansour and S.R. Simon, 1981. Analysis and synthesis of human swing l eg motion during gait and its clinical applications. J. Biomechanics, 14: 823-832. DOI: 10.1016/0021-9290(81)90010-5. PMID: 7328089. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-4BYSFKJ-Y&_user=10&_coverDate=12%2F31%2F1981&_alid=1072088703&_rdoc=1&_fmt=high&_orig=search&_cdi=5074&_sort=r&_docanchor=&view=c&_ct=57&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=3895d250c6f785af7e23a43f37760adb.

Mochon, S. and T.A. McMahon, 1980. Ballistic walking. J. Biomech., 13: 49- 57. DOI: 10.1016/0021-9290(80)90007-X. PMID: 7354094. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-4BYSGGT-DT&_user=10&_coverDate=12%2F31%2F1980&_alid=1072093637&_rdoc=2&_fmt=high&_orig=search&_cdi=5074&_sort=r&_docanchor=&view=c&_ct=79&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=0776c1e1ba6c1818721c33d97ab673a8.

Neptune, R.R., K. Sasaki and S.A. Kautz, 2008. The effect of walking speed on muscle function and mechanical energetic. Gait and Posture, 28: 135-143. DOI: 10.1016/j.gaitpost.2007.11.004. PMID: 18158246. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T6Y-4RFD3MN-1&_user=10&_coverDate=07%2F31%2F2008&_alid=1072141864&_rdoc=7&_fmt=high&_orig=search&_cdi=5043&_docanchor=&view=c&_ct=704&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=b406511df54337fd6ffeba17677a8cea.

Perry, J., 1992. Gait Analysis. SLACK, Thorofare, NJ., pp: 49-167. ISBN-10: 1556421923.

Piazza, S.J. and S.L. Delpt, 1996. The influence of muscles on knee flexion during the swing phase of gait. J. Biomech., 29(6): 723-733. DOI: 10.1016/0021-9290(95)00144-1. PMID: 9147969. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-3VXJG18-3&_user=10&_coverDate=06%2F30%2F1996&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=156a02e548e0135c06b1148f2b38a337.

Riley, P.O. and D.C. Kerrigan, 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. DOI: 10.1016/S0021-9290(98)00107-9. PMID: 7354094. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-3W1H00B-9&_user=10&_coverDate=09%2F01%2F1998&_rdoc=1&_fmt=high&_orig=browse&_sort=d&view=c&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=d9b31ecec9a92e16a02b82d493ee7bc2.

Zarrugh, M.Y. and C.W. Radcliffe, 1976. Simulation of Swing Phase Dynamics in Above- Knee Prostheses. J. Biomech., 9: 283-292. DOI: 10.1016/0021-9290(76)90050-6. PMID: 1270445. http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T82-4BYSG98-B5&_user=10&_coverDate=12%2F 31%2F1976&_alid=1072108347&_rdoc=1&_fmt=high&_orig=search&_cdi=5074&_sort=r&_docanchor=&view=c&_ct=90&_acct=C000050221&_version=1&_urlVersion=0&_userid=10&md5=78689271353d2281b0457f0376b003d2.