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Muscle force distribution during forward and backward locomotion

MICHALINA BŁAŻKIEWICZ*

Józef Piłsudski University of Physical Education in Warsaw, Department of Physiotherapy, Warsaw, Poland.

Backward walking (BW) is a common technique employed in the treatment of a variety of orthopedic and neurological diseases. BW training may offer some benefits especially in balance and motor control ability beyond those experienced through forward walking (FW). The purpose of this study was to determine whether BW represented a simple reversal of FW and, hence muscle force distribution is the same. The study involved one male healthy student of physical education (22 years, h = 185 cm, m = 80 kg). Measurements of spatial-temporal gait parameters were conducted using eight Vicon system cameras, and Kistler plates. Noraxon EMG was used to obtain muscles activity. OpenSim software was used to compute muscle force distribution during both types of gait. During FW and BW there is small difference for force curves produced by m. gluteus maximus (RMS = 0.04), m. biceps femoris short head (RMS = 0.19) and m. tibialis anterior (RMS = 0.16). Good validation by EMG signal was obtained for m. rectus femoris, m. biceps femoris short head, m. tibialis posterior during FW and BW. For m. iliacus, only during BW good validation was achived.

Key words: *muscle force contribution, forward gait, backward gait, OpenSim*

1. Introduction

Many factors such as decline of balance, lean body mass, decrease of muscular strength of lower limbs, weakening of visual, cutaneous, proprioceptive and vestibular senses may lead to falls [1]. Interaction of the sensory system, the motor system and the musculoskeletal system play a great role among all the factors. Wei-Ya and Yan [2] proved that the use of simple physical exercise, such as a backward walking (BW) can improve balance, so it has beneficial effect for healthy individuals. Some motor patterns are easy to describe in terms of e.g. translation and rotation, amplitude, and time. Reversal of direction represents a special kind of movement transformation that may help to get an insight into the internal representations of motor patterns for some classes of movements. The main hypothesis tested by Winter et al. [3] was whether BW could be considered as a simple reversal of forward walking (FW). They suggested that BW was a near image of FW. Grasso et al. [4] found that the kinematic results show simple temporal reversal of joint angles. Vilensky et al. [5] noted that at identical walking speeds, BW was characterized by shorter time of swing and support phase. However, EMG patterns seem to be significantly different between the two movement directions [4], [5]. All simple or difficult movements, that are not habit, require different, not familiar motor control ability. Hence the purpose of this study was to compare muscle force distribution during FW and BW, for person who does not walk backwards as training.

* Corresponding author: Michalina Błażkiewicz, Józef Piłsudski University of Physical Education in Warsaw, Department of Physiotherapy, ul.Marymoncka 34, 00-968 Warsaw, Poland. Tel: +48(22) 8641418, ext. 508, e-mail: michalinablazkiewicz@gmail.com
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2. Materials and methods

2.1. Subject and experimental setup

To quantify contributions of individual muscles during FW and BW one male healthy subject (22 years, h = 185 cm, m = 80 kg) participated in the study. One static calibration trial and one forward and backward gait trial were recorded. The subject was walking along a 10 m walkway at a self select velocity. Kinematic data were collected at 100 Hz using eight Vicon system cameras. Reflective markers were placed using the full body "Plug-in-Gait" marker set. Ground reaction force data were measured at 1000 Hz using two Kistler force plates. EMG activity was recorded at 1000 Hz from six muscles: m. tibialis anterior, m. tibialis posterior, m. rectus femoris, m. biceps femoris short head, m. gluteus maximus and m. iliacus using wireless Noraxon EMG.

2.2. Data processing and comparison method

Kinematic and kinetic data were obtained directly from Vicon. The raw EMG data were exported to MatLab for signal analysis and processing. Raw EMG signals were: firstly high-pass filtered by a Butterworth 10th order filter at 20 Hz, secondly full-wave rectified and thirdly low-pass filtered with a third order Butterworth filter at 5 Hz. The gait EMG processed signals were then interpolated to 100 points per gait cycle. Muscle forces distribution during FW and BW were obtained using OpenSim software [6]. A generic musculoskeletal model with 19 degrees of freedom and 92 Hill-type muscle-tendon actuators was scaled to match the anthropometry of the subject. An inverse kinematics problem was solved to calculate the joint angles of the musculoskeletal model that best reproduces the experimental kinematics of the subject. Inverse dynamics task was solved to determine net moments at each of the joints. To reduce dynamic inconsistencies between the kinematic model and measured ground reaction forces, a residual reduction algorithm (RRA) was used. Following RRA muscle forces were computed using the computed muscle control (CMC) tool. CMC is an optimization-based control technique designed specifically for controlling dynamic models that are actuated by redundant sets of actuators whose force-generating properties may be nonlinear and governed by differential equations. The purpose of (CMC) is to compute a set of muscle excitations that will drive a dynamic musculoskeletal model to track a set of desired kinematics in the presence of applied external forces [7].

In order to compare muscle forces during FW and BW – Root-mean-squared differences (RMS) was calculated between FW and $(\mathrm{BW})^{-1}$ curves

$$RMS_m = \sqrt{\frac{1}{100}\sum_{i=1}^{100}(f_{i,m}(FW) - f_{i,m}^{-1}(BW))^2}\ ,$$

where m – muscle, $f(FW)$ – muscle force curve during forward gait, $f(BW)$ – muscle force curve during backward gait, $f^{-1}(BW)$ – reverse muscle force curve in relation to curve obtained during backward gait. In this paper reverse curve is definite as trajectory in reverse domain $\Omega^{-1} = \{100, 99, 98, ..., 1\}$. RMS value describes magnitude differences between shapes of curves. Only for this paper RMS < 0.38 indicates no difference between the force curves during FW and BW^{-1}. Value 0,38 is average from all obtained RMS values. This criterion depends on the number of muscles used in the computation. In order to compare kinematic and kinetic data and to verify the muscle force distribution during BW and FW, Spearman's rank correlation coefficients were calculated. For that purpose STATISTICA software was used. In this study moderate correlation was noted for coefficient $0.3 \le r_s < 0.5$; strong for $0.5 \le r_s < 0.7$; very strong for $0.7 \le r_s < 0.9$ and almost full correlation for $0.9 \le r_s < 1$ [8].

3. Results

3.1. Kinematic and kinetic data during FW and BW

Figure 1 presents kinematic and kinetic parameters in gait cycle domain during FW and BW. Obtained ankle and knee angles and torques reach lower values during BW in almost whole domain. Moreover, all Spearman's rank correlation coefficients obtained were significant for $p < 0.001$.

For ankle angle correlation coefficient is very strong: 0.7231 and for torque, which in this case was calculated between FW and BW^{-1} curve is: 0.9069. For knee angle: 0.5622 and for torque: 0.7112. For hip angle: 0.8813 and for torque: 0.5582. Obtained results, compared in this way, indicate simple reversal for kinematic and kinetic parameters.

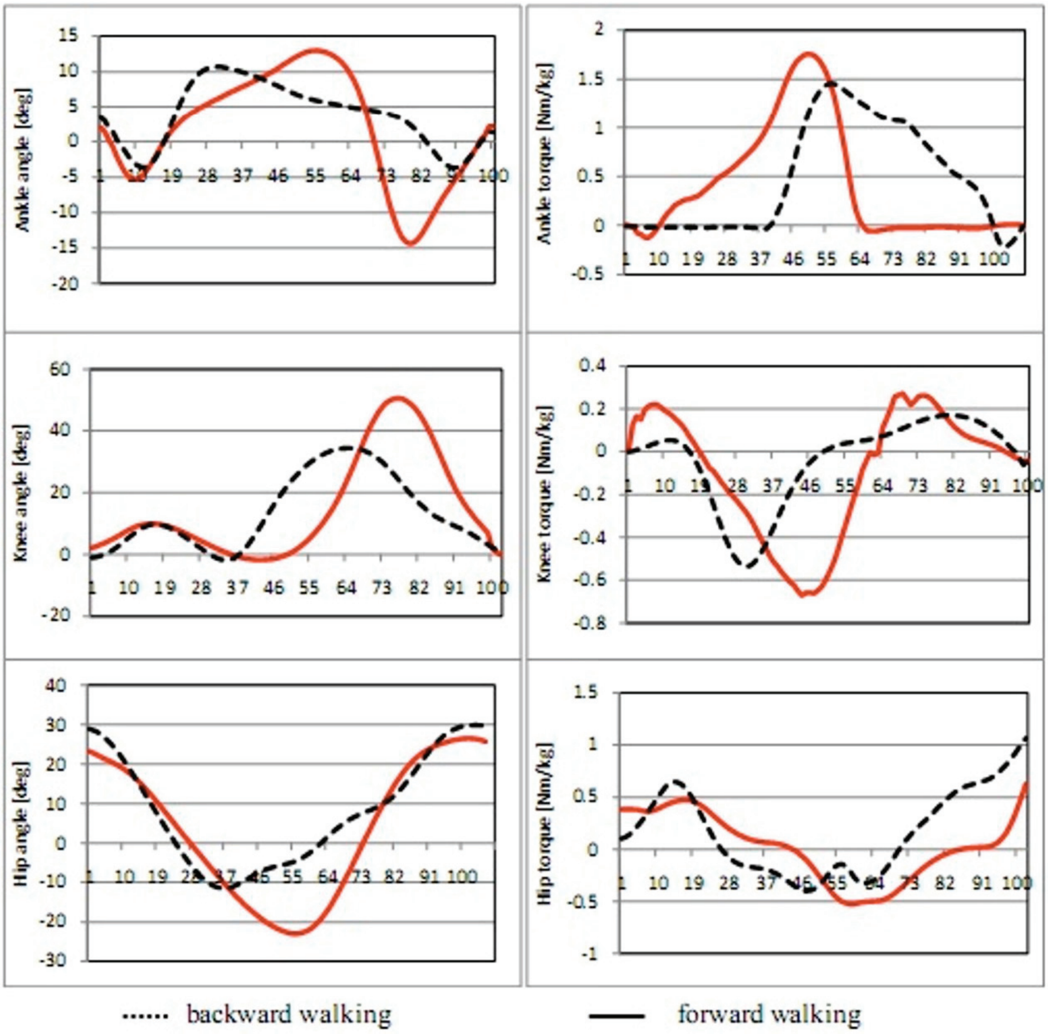

Fig. 1. Kinematic and kinetic parameters for ankle, knee and hip joint during FW and BW

3.2. Comparison of muscle force distribution during FW and BW

Using RMS method for comparison of muscle force distribution during FW and BW it was noted that muscles gluteus maximus and tibialis anterior have reversal force trajectory during BW as evidenced by low values of RMS – 0.04 and 0.16, respectively (Fig. 2).

Muscle rectus femoris chosen as one of knee extensor generates greater force during FW gait and RMS ranged the highest value in the whole set of results: 0.95. While one of its antagonist biceps femoris short head reached RMS – 80% smaller. High RMS values demonstrating a lack of similarity was observed for the other two muscles: tibialis posterior (0.38) and iliacus (0.61).

3.3. Validation of muscle force distribution during FW and BW by EMG signal

Studies of muscle force distributions during FW and BW were validated by EMG signal (Figs. 3–5). In this study, significant – moderate correlation was observed for coefficient $r_s > 0.3656$ for $p < 0.001$. Muscle iliacus was studied as a hip flexor, for which the results are shown in Fig. 3.

For FW, the correlation coefficient between EMG signal and muscle force was 0.0503 and for the same muscle during BW correlation was highest: 0.6452. It means that only during BW for muscle iliacus, there is a significant correlation, so good validation.

For muscle rectus femoris during FW and BW correlation coefficient was at the same level: 0.5663 and 0.5729. For antagonist muscle – biceps femoris short

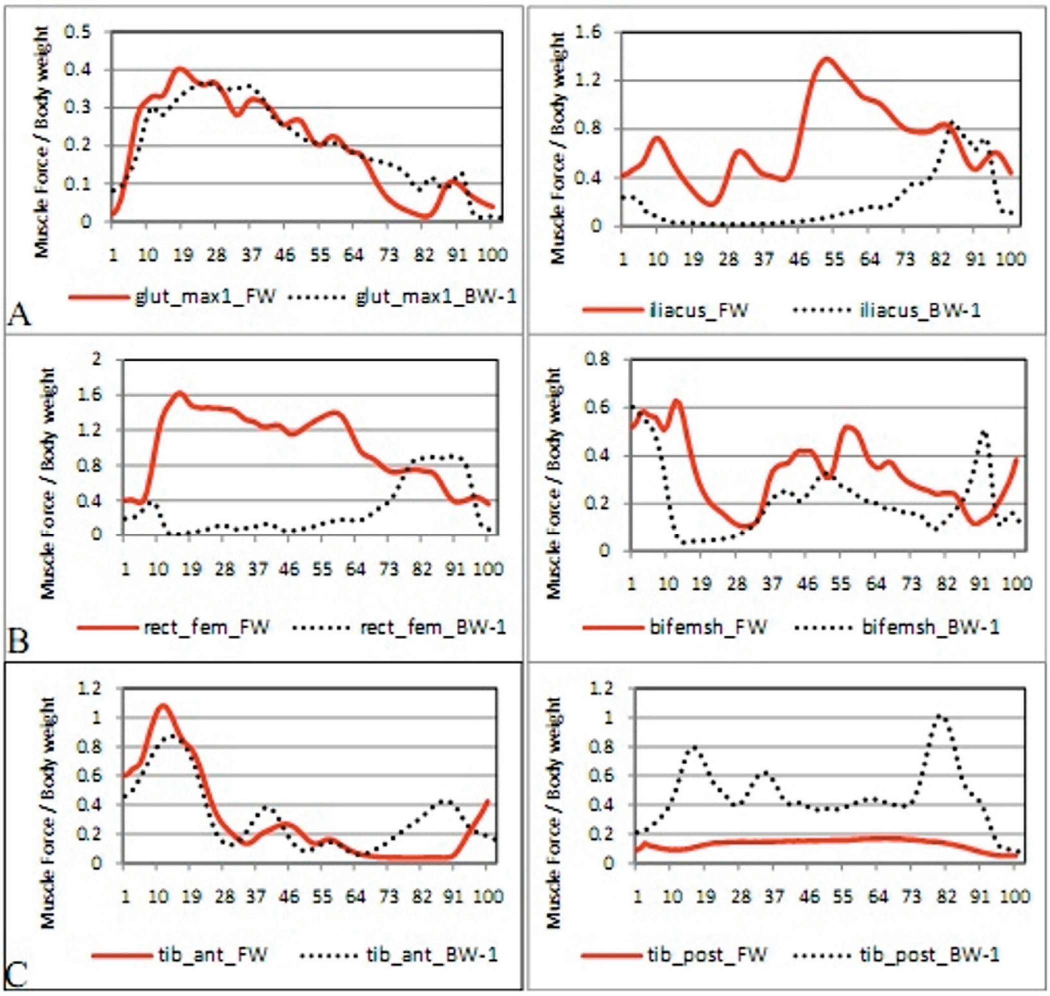

Fig. 2. Muscle force profiles, normalized to body weight during one gait cycle during FW) and BW^{-1}, where: A – hip joint: m. gluteus maximus (glut_max), m. iliacus; B – knee joint: m. rectus femoris (rect_fem), m. biceps femoris short head (bifemsh); C – ankle joint: m. tibialis anterior (tib_ant), m. tibialis posterior (tib_post)

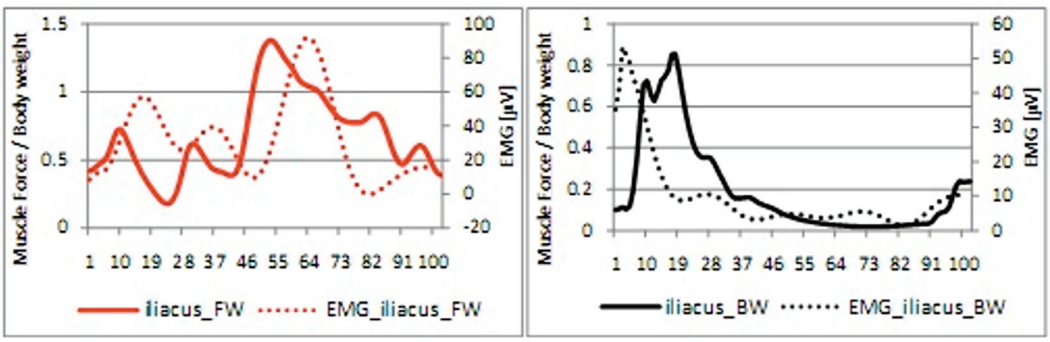

Fig. 3. Muscle iliacus force profiles, normalized to body weight during one gait cycle for FW and BW compared with EMG activity

head the correlation coefficient was lower in comparison with correlation coefficient for m. rectus femoris and was: 0.4126 during FW and 0.4443 during BW. All result for the knee muscles joint were significant ($r_s > 0.3656$ for $p < 0.001$).

For muscle tibialis anterior during FW correlation was 0.3656 and during BW was 0.4924. For antagonist muscle–tibialis posterior correlation coefficient was lower: 0.2081 during FW and 0.0363 during BW. It means that significant correlation is only for muscle

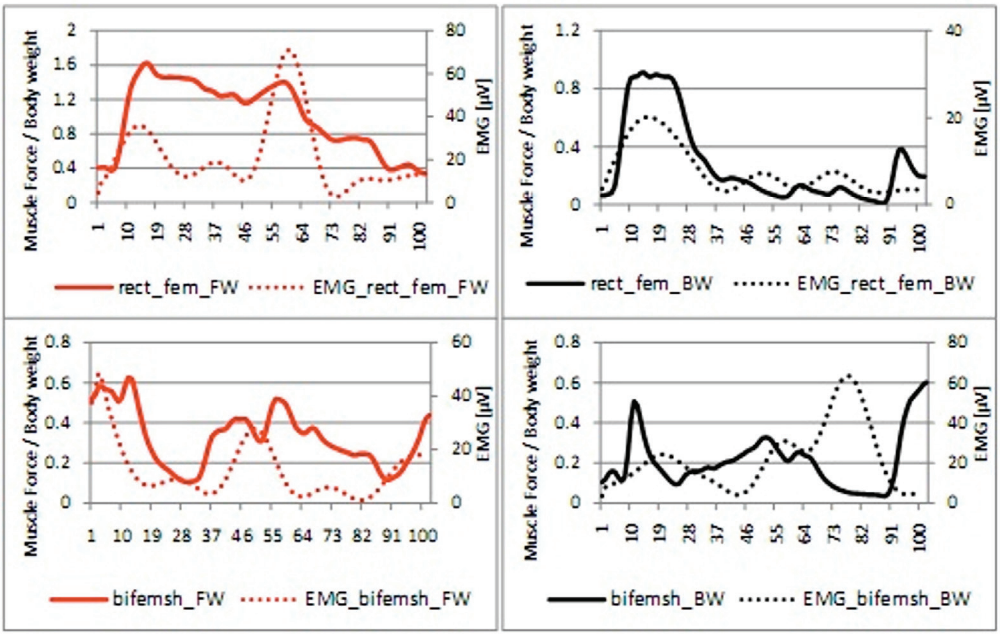

Fig. 4. Muscle recuts femoris and biceps femoris short head force profiles,
normalized to body weight during one gait cycle for FW and BW compared with EMG activity

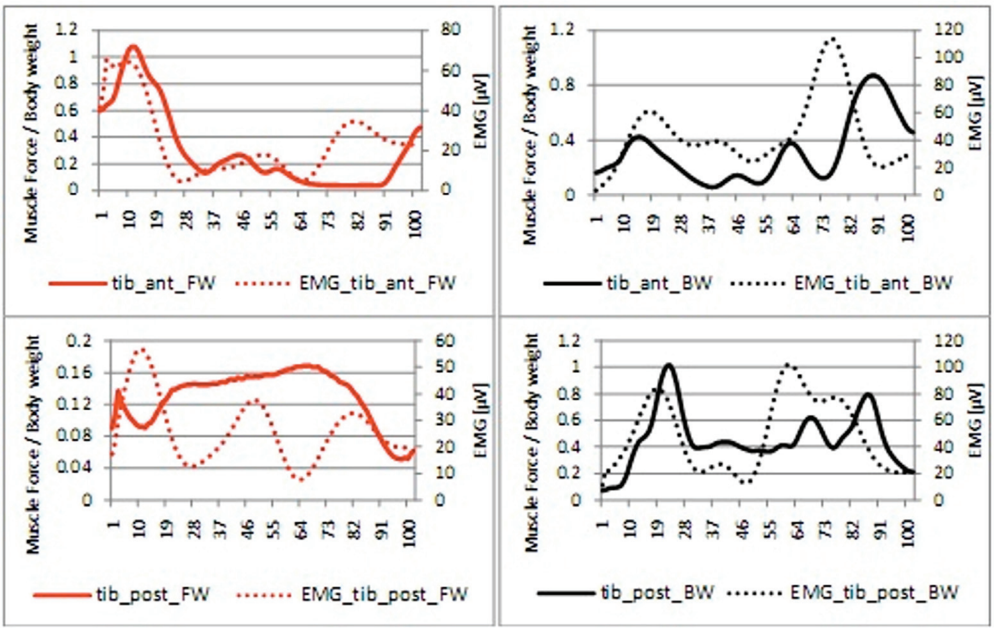

Fig. 5. Muscle tibialis anterior and tibialis posterior force profiles,
normalized to body weight during one gait cycle for FW and BW compared with EMG activity

Table 1. RMS differences and validation of muscle forces during FW and BW by EMG signal

Muscles	Differences	Validation	
	RMS	FW force vs EMG	BW force vs EMG
m. gluteus maximus	0.04*		
m. iliacus	0.61	0.0503	0.6452*
m. rectus femoris	0.95	0.5663*	0.5729*
m. biceps femoris short head	0.19*	0.4126*	0.4443*
m. tibialis anterior	0.16*	0.2081	0.0363
m. tibialis posterior	0.38	0.3656*	0.4924*

* – no differences

tibialis posterior during both movements. Summary of all results is presented in Table 1.

During FW and BW there is no difference for force produced by m. gluteus maximus, m. biceps femoris short head and m. tibialis anterior. Good validation by EMG signal was obtained for m. rectus femoris, m. biceps femoris short head, m. tibialis posterior during FW and BW. For m. iliacus, only during BW a good validation was obtained.

4. Discussion

Walking style represents a complex whose significance encompasses both the mechanical requirements of gait, such as: equilibrium, speed, and energy as well as the emotional aspects of life [9]. During forward and backward walking muscles produce forces which act directly on the skeleton. These forces influence whole body movement, as ground reactions cause the effects of muscle force to be transmitted to segments remotely from the muscular contraction. This effect is referred to as "dynamic coupling". Understanding the complex mechanisms behind normal gait is challenging [10]. The main aim of this study was to quantify the difference in the leg muscle force patterns during FW and BW, and validation of muscle force distribution by EMG signal. However, the relationship between EMG and muscle force is not trivial [11]. To do this, RMS and correlation coefficients were used. Błażkiewicz and Wit [12] proved that correlation coefficients and RMS are good methods in comparison with the closeness of the two curves in the shape. They performed comparative analysis of sensitivity of four methods: waveform parameterization, correlation coefficients, RMS and IAE (Integral Absolute Error) in order to compare joint angles with reference curve. The sample scores obtained in this work provide important information about closeness in the shape of two curves. Moreover, authors encourage using multiple techniques of data analysis.

Regarding the gait analysis, most publications concern the analysis of muscle forces, muscle activity, kinematic and kinetic parameters during simple forward walking [6], [13]. For FW and BW mixed results were reported. Grasso et al. [4] suggested that the preservation of kinematic templates for the motion of the joints between BW and FW was reflective. In the present study it was proved that changes of hip angle during BW tend to be the time reversed of those during FW, but the extent of FW–BW correspondence was smaller for the knee angle and ankle angle. The correlation between the joint torques in two movement directions was inhomogeneous. Very strong and almost full correlation was observed for the knee torque (0.7112) and the ankle torque (0.9069) and strong correlation for the hip torque (0.5582). The same results for kinematic and kinetic parameters were obtained by [3]–[5]. Furthermore, Winter et al. [3] calculated that correlation between joint powers was generally lower than that of the corresponding joint angle or torque. The EMG patterns analyzed by Grasso et al. [4] were different in these two movement directions. Authors argued that conservation of kinematic parameters across gait reversal at the expense of a reorganization of muscle synergies does not arise from biomechanical constraints but may reflect a behavioral goal achieved by the central networks involved in the control of locomotion.

Individual muscle forces evaluated from experimental motion analysis may be useful in mathematical simulation, but require additional musculoskeletal and mathematical modeling. In this study, OpenSim with numerical method of optimization was used to evaluate muscular forces during gait. The developed optimization method calculates optimal forces during gait, given a specific performance criterion, using kinematics and kinetics from gait analysis together with muscle architectural data. Measuring exact muscle forces during gait analysis is not currently possible. Experimental methods validating mathematical methods to calculate forces are limited. Electromyography (EMG) is frequently used as a tool to determine muscle activation in experimental studies on human motion. A method of estimating force from the EMG signal, the EMG-to-force approach was recently developed by [14] and is based on normalization of activation during a maximum voluntary contraction to documented maximal muscle strength. In this work, muscle forces obtained from OpenSim show reasonably good correlation with EMG signal in the knee flexor and extensor muscles, but less correlation in the plantarflexor and dorsiflexor muscles. The results presented for FW are similar to those obtained by [15].

The mechanics of BW is different from the one of FW. Stance phase during gait begins with heel strike and ends at toe-off in FW. On the contrary, in BW toes contact the ground first and the heel is lifted off the ground at the end. The foot impact on the ground in early stance is sustained by co-activation of several limb muscles: flexors and extensors at the hip, knee, and ankle in FW, whereas the same event is accompanied by activity in knee extensors and ankle plantarflexors in BW. Thus knee flexors tend to be reciprocally activated with knee extensors in BW gait, whereas they are roughly co-activated in FW gait. Conversely, ankle flexors tend to be reciprocally activated with ankle extensors in FW gait, whereas they are co-activated for extensive periods of BW gait cycle. The magnitude of EMG activity integrated over one gait cycle generally is greater in BW gait than in FW gait [3]. This finding suggests a greater level of energy expenditure in BW than in FW and a greater level of oxygen consumption in BW than in FW, which was reported by [16], [17]. Moreover, the anatomic and functional asymmetry of a foot and a leg along the anterio-posterior axis also imposes different biomechanical constraints on BW and FW gait. Calf and thigh muscles are highly asymmetrical about the frontal plane; the mass and strength of the muscles on the posterior aspect of the calf (m. triceps surae) and on the anterior aspect of the thigh (m. rectus femoris) are much greater than those of the muscles on a respective opposite side. All these asymmetries may well explain the lack of great correspondence of many gait parameters between FW and BW directions [4]. In fact, in this paper lack of similarity for muscles force trajectory during FW and BW was observed for muscles: tibialis posterior, rectus femoris and iliacus. However, quite good agreement is observed for their antagonists. Muscle force distribution during BW is not a simple reversal of those obtained during FW. Because BW is not a common movement, so it was performed more carefully. Probably the result obtained can be explained by different reorganization of the muscle synergies or neuromotor control in lower limbs during BW. Obviously, this hypothesis requires verification in a study with a larger number of subjects.

Acknowledgements

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