Does Really Lower Limbs Behave Asymmetric in Gait of People Without Impairments?

Heydar Sadeghi

Associate Professor, Department of Physical Education & Sport Sciences, Tehran Tarbiat Moallem University

Abstract

Using two consecutive gait cycles, simultaneous and bilateral kinetic gait data, the main objectives of this study were a) to identify the main functional roles of ankle, knee and hip extensors/flexors, and b) to determine whether the action taken by these muscle groups appears to be symmetric or not. Gait of our able-bodied subjects appears to be asymmetric with significant differences noted between each two corresponding peak muscle moment values. Using Principal Component Analysis (PCA) as a curve structure detection method, task discrepancies were recognized when comparisons were made between each two corresponding representative moment curves at each joint (local asymmetry). Muscle moment behaved symmetrically when the right limb representative curve was compared to its corresponding principal component (PC) at the contralateral limb. Gait of able-bodied subjects appears to be symmetric, while control and propulsion were recognized as two major roles of the extensors and flexors (global gait asymmetry). Symmetrical behavior of the lower limbs should be considered a consequence of local asymmetry which indicates different levels of within and between muscle activities developed at each joint during gait cycles.

Key words: Biomechanics, Gait, Symmetry, and Principal Component Analysis.

Introduction

Because pathological conditions can affect gait [1-4], understanding the fundamental tasks of the lower limbs in able-bodied subjects can guide clinicians in refining their clinical evaluation or rehabilitation treatment.

Although the issue of whether the lower limbs behave symmetrically or not in the gait of people without impairments is still debatable [5], a few assumptions have been proposed to explain symmetry between the lower limbs. These include the effect of limb dominance [6-8], different levels of muscle contributions in gait to achieve control and propulsion [9-11] and within- and between-limb compensatory mechanisms [2,12,13]. To our knowledge, none of these assumptions have been objectively evaluated. Studies related to gait symmetry were often limited since the analysis was performed on a single lower limb at a time [14] rather than collecting simultaneous bilateral data. Discrete values such as peak or zero crossing were also used [11,15] and analyzed instead of using complete curve information. Though these methods of gait analysis provide some information, it would be reasonable to claim that bilateral gait pattern analysis is more appropriate to test the symmetry hypothesis.

This study was performed to test the hypothesis that lower limb symmetry is present in able-bodied subjects. It was postulated that the gait of people without impairments is symmetrical if the net muscle moments developed at the ankle, knee and hip of the right and left lower limbs during two consecutive gait cycles are similar. Muscle moment was chosen since it provides valuable insight into the mechanical causes of movement introduced by agonist and antagonist muscles as an integration of all the neural controls acting at each joint [16]. Since most of the functional activities occur in the plane of progression during the stance phase, the muscle moments in the sagittal plane were chosen. Principal Component Analysis (PCA) was applied as a multivariate statistical method which has the capability of detecting and classifying the main structure of the curve data [17]. In short, this gait study was undertaken a) to identify the main actions taken by the ankle, knee and hip extensors/flexors and overall lower limb muscle activity during able-bodied gait, and b) to determine whether these actions appear to be symmetrical or not.

Methods

Sixty gait trials were obtained from 20 healthy male subjects having an average age of 25.3 ±4.1 years, height of 1.77±0.06 m and average mass was 80.6 ±13.8 kg. They had no previous history of either orthopedic or neurological ailments, such as a recent injury or surgery, which could affect their walking pattern. Subjects received a stipend to cover their travel expenses and time.

A three-dimensional (3D) seven segment model consisting of the trunk, thighs, shanks and feet was defined using twenty reflective markers with a diameter of 2.5 cm. The model and the procedure has been explained in detail elsewhere [13]. In summary, markers were placed over the lateral malleolus, heel and lateral border of the fifth metatarso-phalangeal joint for each foot while markers placed over the apex of the lateral epicondyle and the mid-lateral side of the tibias located the shanks. Markers were also placed over the mid-lateral side of the thighs and the greater trochanter to define the thighs. For the pelvis, markers were placed over the anterior superior iliac spines and crests of ilium. The pelvic markers as well as markers placed over the lateral border of the shoulders identified the trunk. Measurements were taken between the external markers and the estimated joint center of rotation of each lower limb to calculate motion in the joint coordinate system.

Bilateral gait data were collected with an eight video-based camera system* (90 Hz) synchronized to two AMTI force plates= (360 Hz). Four cameras were placed on either side of the subject at an average distance of 4.5 m and located along an arc of about 120° to cover two consecutive strides. Subjects were asked to walk at a self-determined pace along a 13m walkway. The walkway was designed to allow the subjects to walk comfortably and to make contact with the force plates. The right limb was always the leading limb [13]. The best three trials out of five were selected. These corresponded to the trials where the right and left feet made contact with the first and second force plates, respectively.

Direct Linear Transformation software from the Motion Analysis Expert Vision system was used to reconstruct the image markers into three-dimensional coordinates. A fourth order zero-phase lag Butterworth low-pass filter was applied to reduce the noise in the video data. The cut-off frequency was 6 Hz for the body segments and 30 Hz for the force data. For averaging purposes, moments were normalized with respect to body mass. Data could also be normalized based on the height of the subjects. However, since our subjects were more or less the same height, it was assumed that it would not affect the outcome much. Data were further normalized with respect to the duration of the gait cycle (GC) of each limb. Joint moments were expressed according to the convention proposed by the International Society of Biomechanics and included in Winter [18], where the extensor and plantarflexor moments are considered positive. A gait cycle corresponds to the period beginning with foot contact and ending with the following heel contact of the same limb. The mean stance phase duration was 60.7% GC for the right and 61.0% for the left limb.

---

* Motion Analysis Corp., 3617 Westwind Blvd., Santa Rosa, CA 95403, USA.

= Advanced Mechanical Technology Inc., 176 Waltham St., Watertown, MA 02472, USA.

Kinematic and force plate data were used in an inverse dynamic approach to calculate the net sagittal muscle moments at the hip, knee and ankle of the lower limbs during the stance phase. It is important to note that the term 'moment' was used throughout this study to express the resultant effect of the forces exerted by the muscles crossing the joint [19]. The moment bursts were labeled using an alpha-numeric code where the letter refers to the joint and the number indicates the sequence of the moment burst. For example, A1 corresponds to the first peak moment of the ankle.

Using the average value along the curve, the variation in the moment curve developed at each joint during the stance phase was estimated by the coefficient of variation (CV%) to provide an overall impression of the variability in the data. Student's t-test for paired data with a $p<0.05$ threshold was performed on the right and left limb peak muscle moments as a primary evaluation of limb symmetry.

PCA was applied to identify the main structure of the data throughout the variation in the data. A principal component analysis was performed twice during able-bodied gait: once to identify the actions of each joint separately and a second time to determine symmetry between the lower limbs by simultaneously analyzing all the joints of each limb. Lower limb symmetry was assumed to be present if the Principal Component (PC) curves derived from each joint or from each of the lower limbs described the same portion of the stance phase. Four steps were involved in the PCA application. The first step consisted of finding the covariance matrix of the muscle moment curves during the stance phase of the gait cycle. A matrix was created from the muscle moment data of a specific joint or joints of a lower limb. It consisted of 60 rows, each row representing a single trial of a subject (20 subjects X 3 trials) and 61 columns which contained the instantaneous muscle moment of that joint during the stance phase calculated at each percent of the gait cycle (0 to 61%). For the simultaneous analysis of all the joints of a lower limb, a second matrix was formed and consisted of 180 rows and 61 columns. The first sixty rows contained the ankle data of each subject (20 subjects X 3 trials), while the remaining 120 rows contained the knee and hip data of the same limb. The 61 columns represent the instantaneous moments for the stance phase calculated at each percent of the gait cycle.

The purpose of applying the PCA is to extract the maximum variance from the data by means of a few orthogonal components called principal components (PC). The first PC is the linear combination of the observed variables that maximally separate the subjects by maximizing the variance of their component scores. In our study, each PC contained 61 values each having a factor loading. When plotted against the stance phase duration, as shown in Fig. 2.3, a factor loading curve was obtained. This was called the PC curve. The second step was to choose the number of PCs which should be retained for further analysis. The eigenvalues of each PC indicated how many components are important in conveying most of the major information. Here the first two PCs which accounted for over 80% of the variance on average were kept. The third step was to choose and perform an appropriate type of rotation on the PCs to maximize the variation leading to more interpretable physiological information.[20] The Varimax rotation was used to rotate the PC axes. The last step was to give a physical meaning to each PC. Names were given to the PCs according to what each representative curve describes in terms of muscle activity during the stance phase. To determine what each PC measures, the muscle moment having the highest correlation within each PC (called the factor loading) was used. In this instance, a factor loading higher than 0.70 was used for further biomechanical interpretation [21]. Lower limb symmetry is thought to be present if each two corresponding representative curve (PCs) derived from the right and left limbs describe the same portion of the stance phase with a loading factor of 0.7 or over [22].

We proposed that the role of the muscles could be identified using PCA. We presumed that gait symmetry between two corresponding lower limb joints could be quantified by means of the PC curves if the significant factor loading (0.7 or over) was similarly distributed (local gait symmetry) in the stance phase. For the lower limbs, gait symmetry could be assumed if the corresponding PCs derived from all the lower limb joints described the same portion of the stance phase (global gait symmetry).

Results

The average sagittal muscle moment curves and their standard deviation developed at the right and left ankles, knees and hips during the stance phase are presented in Figure 1.

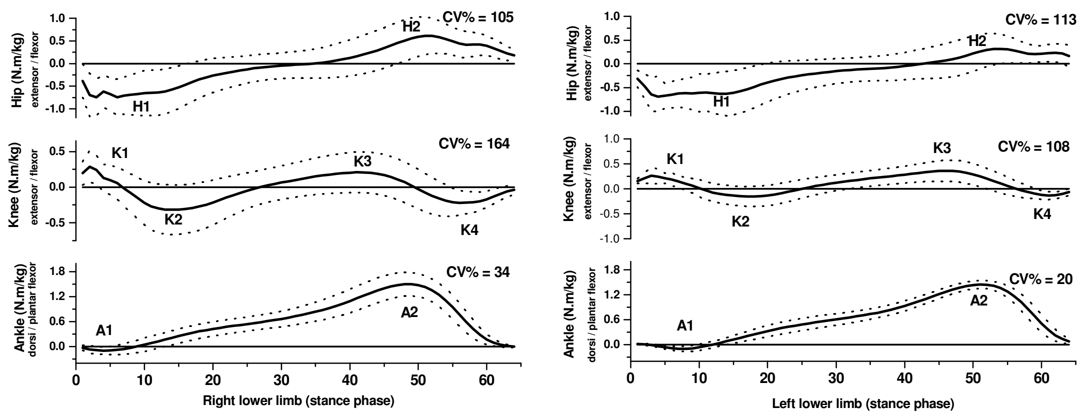

Figure 1: Average sagittal muscle moment curves and standard deviation (1SD) developed at the ankle, knee and hip joints during the stance phase of 20 able-bodied young male subjects

Muscle moment curves reported in this study were in close agreement in shape and magnitude with previously published findings [1,23,24]. The coefficient of variation (CV) varied between 34% and 164% with an average of 103% for the right limb, while the CV for the left limb ranged between 20% and 113% with a mean of 80%. The CV value at the right limb was 21% greater on average than the left limb. The greatest within-limb variations occurred at the right knee (CV = 164%) and left hip (CV = 113%), while the lowest within-muscle activity variation was noted at the left ankle (CV = 20%). The peak moment values are given in Table 1. The right limb peak values were greater than the left limb values by 11% on average.

The terms given by Adams and Perry [25] were used to present the functional tasks, gait phases and duration of the phases in the stance phase. According to their definition, three functional tasks are recognizable, namely weight acceptance, which includes both loading response and initial contact (0-10% of GC), single limb support which includes mid-stance (10-30% of GC) and terminal support (30-50% of GC), and limb advancement which includes propulsion (50-60% of GC).

The typical ankle muscle moment curve shows a short period of activity by the dorsiflexors (A1) which usually occurs in the first 10% of the gait cycle (GC) to control the lowering of the foot [26]. Then the plantarflexor moment (A2) increases and dominates from 10% to about 50% of GC [27] to resist and control the forward rotation of the tibia over the foot [26,28,29]. Late stance is the period that has been the focus of a major controversy regarding the functional role of the ankle plantarflexors. While some authors agree that the ankle plantarflexors are the main source of energy to propel the trunk upward and forward after heel-off, and in many cases until toe-off [30-33], the entire push-off concept has been questioned by others [34-38] who believe that the ankle plantarflexors function to restrain, not accelerate, the trunk over the ankle in walking. The ankle plantarflexors have also been characterized as an accelerator which facilitates the movement of the leg into the swing phase [39-41].

Table 1: Peak muscle moments and standard deviation (SD) values calculated at the ankles, knees and hips (* $p < 0.05$) for 20 healthy young male subjects

Joint	Peak	Right		Left	
		Mean	**SD**	**Mean**	**SD**
Ankle	**A1**	-0.15	0.09	-0.12	0.02
	A2	1.53	0.30	1.46	0.10
Knee	**K1**	0.27	0.17	0.28	0.10
	K2	-0.40*	0.25	-0.26*	0.15
	K3	0.26	0.24	0.36	0.22
	K4	-0.31*	0.14	-0.20*	0.13
Hip	**H1**	-0.83	0.34	0.83	0.35
	H2	0.77*	0.32	0.40*	0.24

The limb enters the stance phase while the sagittal knee moment goes to net extension with a momentary activity taken by the knee flexors during the first few percent (0-10%) of the gait cycle (K1). The stance phase knee flexion occurs as a shock absorber where the knee flexors contribute to weight acceptance and also to effectively shorten limb length and prevent excessive vertical translation of the body center of mass [42]. Afterwards, the extensors (K2) are involved to slow the knee flexion in mid-stance as full weight bearing takes place (about 10-30% of GC). During the terminal stance a flexor moment (K3) is evident (30-50% of GC) creating an extension force and bringing the knee joint into extension by midstance. It should be noted that this passive extension could not occur without the strong eccentric contraction of the plantar flexors restraining the shank from progressive forward rotation. Just after toe-off, the knee extensors (K4) activate to decelerate the backward rotating leg and minimize heel rise (at about 65% stride) [43].

During the stance phase, the hip extensor activity (H1) which occurs shortly after heel-strike has been associated by other researchers with control of the forward acceleration of the trunk [26] and the potential collapse of the stance limb[24] as well as with forward progression [12,13,26]. The hip flexors (35% GC) then dominate to support the body weight transfer and maintain body balance during the mid-stance period. Winter et al. [26] further suggested that from 50 to 70% of GC, the hip flexors (H2) are responsible for pulling the lower limb up and accelerating the thigh and leg forward prior to and shortly after toe-off.

The eigenvalues related to the variance of the moment data extracted by each PC are presented in Table 2. The third and highest PCs which accounted for the remaining variations were not taken into consideration since they presented random variations [15,20] which are difficult to interpret. The first two representative curves (PCs) which accounted on average for over 80% of the information in the original moment curves developed at the right and left ankle, knee and hip are presented in Figure 2. The first PC accounted for 51% and 40% of the variation in the right and left ankle moment curves (Table 2), while the significant factor loading (over 0.7) was distributed between 18 to 35% (right limb) and 15 to 35% (left limb) of the GC. Both corresponding PCs mostly highlighted the contribution of the ankle muscle during single support. The second PC for the right and left ankles accounted for 21% and 20% of the variation while the significant factor loading was distributed between 50 to 60% and between 35 to 50% of the GC. The right ankle moment highlighted muscle activity during the propulsion phase while the left ankle moment (PC2) described the muscle activity occurring at the terminal stance.

Table 2: The variance extracted by each PC from the right and left lower limb muscle moment data

Joint	% Right lower limb			% Left lower limb		
	PC1	PC2	Total extracted variation	PC1	PC2	Total extracted variation
Ankle	51	21	72	40	20	60
Knee	80	11	91	73	12	85
Hip	73	15	88	82	6	88
Lower limb	64	29	93	63	30	93

At the knees (PC1), the significant factor loading spread out between 5 to 55% of the GC for the right limb and 15 to 57% of the GC for the left limb, and accounted for 80% and 73% of the total variance. The right limb knee muscle activity highlighted the role of the knee muscle during weight acceptance, single limb support and partially in propulsion, while the corresponding PCs at the left limb described the role of the knee muscle during single limb support and partially at the propulsion phase. PC2 for the knees described 11% (right) and 12% (left) of the variation, and the significant factor loading was distributed over the first 5% (right) and between 2 to 10% (left) of the GC. The right knee muscle activity corresponded to the beginning of the initial contact period, while the left knee muscle moment highlighted the muscle activity during almost the entire initial contact period.

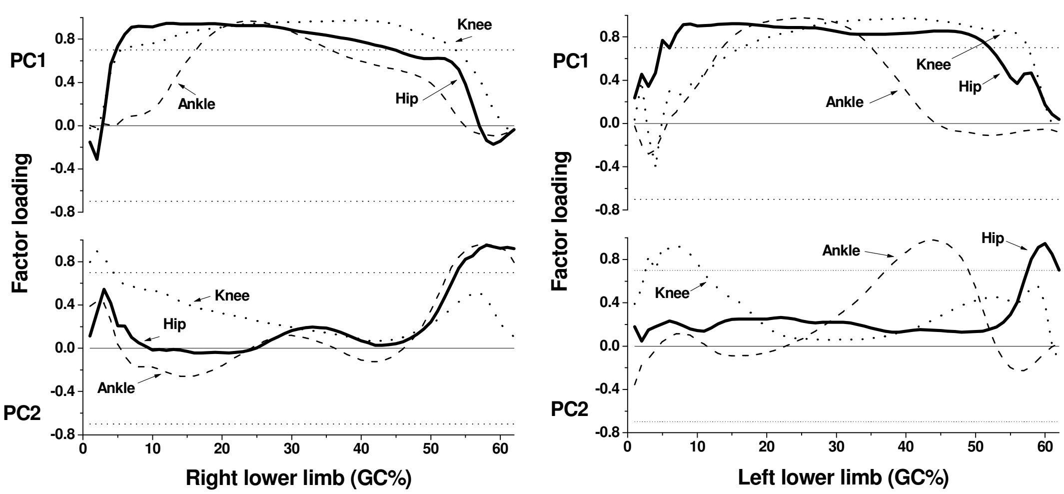

Figure 2: The first two PCs extracted from muscle moment curves calculated at the right and left ankles, knees and hips

PC1 at the right and left hips accounted for 73% and 82% of the data variation, respectively, and their significant factor loading was distributed between 5 to 45% (right limb) and 5 to 52% (left limb) of the GC. Weight acceptance and single limb support were the common portion of the stance phase highlighted by the hip sagittal moment (PC1) while the beginning of the propulsion phase was also highlighted by the right hip moment. For the right hip, PC2 described 15% of the data variation, while the highest factor loading was distributed over 50 to 60% of the GC. The corresponding component (PC2) for the left hip accounted for 6% of the variation with a significant factor loading distributed between 55 to 62% of the GC. The right hip moment highlighted the propulsion phase, while the left hip muscle activity described the second part of the propulsion phase.

Figure 3 presents the first two representative curves (PCs) from the PCA applied over the whole muscle moment curve calculated for the right and left lower limbs. The first PC1 accounted for 64% and 63% of the data variation of the right and left limbs respectively. The significant factor loading (0.7 or over) was distributed between 20 to 60% of the GC in both the first corresponding PCs. These PCs highlighted the role of sagittal muscle activity at the lower limb during the mid-stance and propulsion phases. The significant loading factor for the second PC for both lower limbs spread out from the first 20% of the GC to the end of the stance phase, describing 29% and 30% of the data variation for the right and left limbs, respectively. These PCs describe the role of muscle activity at the lower limbs during weight acceptance and loading response.

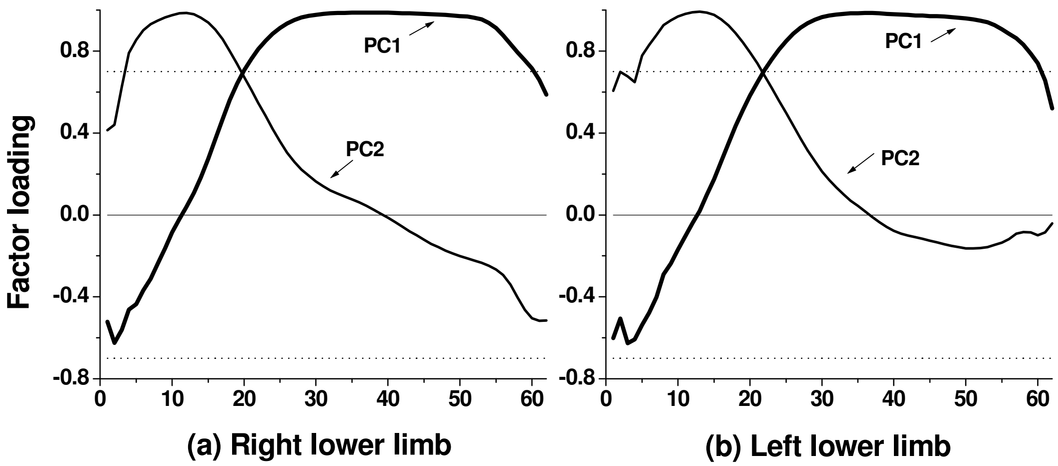

Figure 3: The first two PCs for the muscle moments developed at the lower limb during the stance phase of 20 healthy young male subjects

Discussion

The purpose of this bilateral gait study on subjects without impairments was to identify the roles of the sagittal plane joint moments to characterize local and global symmetry. In the literature, gait symmetry was assumed when no statistical differences in the t-test, anovas, etc. were reported between two corresponding parameters [11,12,44]. Using peak moments only, joint or local asymmetry was observed in our able-bodied subjects during the stance phase except for the peak knee extensors (K1) and flexors (K3). Using a single parameter and most importantly discrete values could be an initial step to describe the function of a specific joint during gait and determine if its muscle activity is within a normal range or not. But to characterize entire lower limb behavior and determine if gait is symmetrical or not, it is essential to apply a multivariate analysis method such as PCA which has the capability of detecting the structure of the data.

The first objective of this bilateral gait study on able-bodied subjects was to identify the roles of the sagittal plane joint moments taken independently to characterize local joint symmetry. To achieve this objective, we proposed that gait symmetry can be quantified by means of the PC curves. For a specific joint (local symmetry), we suggest gait symmetry could be assumed if the corresponding PCs of the right and left values highlighted the same portion of the stance phase.

Using the above working assumption, no significant differences were noted for the ankle's first identified functional task (PC1), suggesting local symmetry. During this period (15 to 35%) of the gait cycle, the plantarflexors eccentrically contracted to control and maintain balance of the center of mass [45] while the plantarflexors restrained the tibia during forward progression over the foot. This observation was in contrast to Winter et al.'s [46] finding that the ankle muscles do not contribute to dynamic balance during the stance phase. However, it was somewhat similar to that of Kepple et al. [47] who stated that the ankle plantarflexors are the major source of preventing the collapse of the upper body and supporting the body weight transfer during stance. Differences, however, were noted for the ankle's second role (PC2) where a significant factor loading of 0.7 or over was observed at a different portion of the stance phase for the right (50-60% of the GC, propulsion) and left limb (35-50% of the GC, terminal stance). This indicated local asymmetry for the second identified task (PC2) for ankle muscle activity during the stance phase.

Local asymmetry was found at both knee and hip levels. The knee muscle activity (PC1) was observed at initial contact and at the beginning of mid-stance (5-15% GC) to accept the weight, stabilize the pelvis and decelerate the mass. The role of the right knee moment (PC2) was identified as stabilizing weight-bearing, while for the left knee (PC2), the role of the muscles was to absorb the impact of the dropping body on the limb [48].

A significant discrepancy was noted at the end of the terminal stance (45-52% of the GC) between right and left hip sagittal muscle activity (PC1). Winter [18] used the term 'dynamic balance' for the actions taken by the muscle moments during 5 to 45% of the gait cycle. These muscles have also been characterized as a major source of controlling the H.A.T. and center of mass movement [49]. Applying the above biomechanical information, PC1 for the right hip seems to explain the control balance function whereas for the left hip muscle moment (PC1), the significant factor loadings were distributed between 5 to 50% of GC, illustrating muscle activity in both single (control balance) and double (between limb coordination) limb support periods.

For the right limb, the second PC for the ankle and hip moments accounted for 21% and 15% of the variation, respectively. The significant factor loading values on these PCs (Figure 2) were attributed to both ankle plantarflexors and hip flexors acting during the propulsion phase (50-60% of GC) [11,30]. Though the role of the ankle plantarflexors was not highlighted by the first two PCs for the left limb, the importance of the action taken by this group of muscles was shown by the second PC for the right limb. Furthermore, based on the results obtained from this study, though the role of the ankle plantarflexors in the propulsion period (50-60% of GC) was not highlighted as the first major ankle muscle activity as suggested by Winter [33], a passive contribution cannot be an appropriate characteristic for the ankle plantarflexors [36]. Moreover, the role of the hip flexors (PC2) in propelling the body weight forward was highlighted while the muscles contracted concentrically. Our result is in agreement with previous studies [11,13,49] which reported that during 50 to 60% of the gait cycle, the hip flexors were mainly responsible for propelling the body forward in the plane of progression by pulling the thigh up and forward. However, it is important to note that hip contribution to propulsion was recognized as the second functional task of the hip flexor moments in this study.

Our results showed both ankle plantarflexors (49% of GC) and hip flexors (51% of GC) reached their maximum magnitude when ankle plantarflexion (47% of GC) and hip flexion (51% of GC) were at their maximum angular position. These observations might provide insight into coactivity and some sort of explanation regarding the idea of complementary activity between muscles acting at the hip and ankle. Earlier onset and longer distribution of higher factor loadings (over 0.7) on the hip (PC1: 5 to 50% of GC) compared to the ankle plantarflexors (PC1: 15 to 35% of GC) might lead us to draw conclusions about the secondary or complementary role of the ankle plantarflexors compared to the hip extensors/flexors. Continuation of the functional contribution of the hip flexors (35 to 40%) might also be explained by the peak muscle moment at the hip level that was observed (51% of GC) later than the ankle plantarflexors peak (49% of

GC). This result could be further considered as a unique stance phase task for the action taken by the muscles at the ankle and hip particularly during single limb support (15 to 35% of GC).

The second objective of this bilateral gait study was to identify the role of the sagittal plane joint moments taken together to characterize global gait symmetry. The two first representative curves (PC1 and PC2) accounted for the largest and an almost equal proportion of the observed variables' variance for the right (93%) and left (93%) limbs in the sagittal plane. In both PC1 and PC2, the significant loading factor values were similarly distributed over 20 to 40% and 5 to 20% of the gait cycles. These results might explain in part the idea of gait symmetry (global), while discrepancies were noted for group of muscles acting at each two corresponding joints (local). It seems that compensatory mechanisms might be the best explanation to describe global gait symmetry while different actions are taken by the joints.

This study was designed to identify the roles of the actions taken by the ankle, knee and hip muscles in the sagittal plane in a young able-bodied population. This could be useful in understanding the differences between the gait of healthy and pathological subjects. The PCA method provides specific information on the population sampled. It is not known if the results obtained from one population can be safely applied to a completely different population. The balance, control and propulsion characteristics should be present but not necessarily with the same intensity.

Conclusion

PCA was able to identify the two main functional contributions of ankle, knee and hip sagittal muscle moments during the gait of subjects without impairments for identifying local and global symmetry. Local asymmetry in the gait of people without impairment is suggested, based on different functional tasks between the right and left hips, knees and ankles to control balance, between limb coordination and propulsion functions. The lower limbs, on the other hand, appeared to behave symmetrically when the total behavior of the limbs is considered. Compensation is recognized as an explanation for the existence of local asymmetry.

References

1. Crowinshield RD, Brand RA, Johnston RC. The effects of walking velocity on hip kinematics and kinetics. Clin Orthop 1978; 132:140-44.
2. Winter DA, Sienko SE. Biomechanics of below knee amputee gait. J Biomech 1989; 21:361-67.
3. Gitter A, Czerniecki J, Meinders M. Effect of prosthetic mass on swing phase work during above-knee amputee ambulation. Am J Phys Med & Rehabil 1997; 76(2):114-120.
4. Sadeghi H, Allard P, Duhaime M. Muscle power compensatory mechanisms in below-knee amputee gait. Am J Phys Med Rehabil 2001; 80: 25-32
5. Sadeghi H, Allard P, Prince F, Labelle H. Gait symmetry and limb dominance in able-bodied gait. Gait & Posture 2000; 12(1): 34-45.
6. Hirasawa Y. An observation on standing ability of Japanese males and females. J Anthrop Soc Nippon 1979; 87: 81-92.
7. Õunpuu S, Winter DA. Bilateral electromyographical analysis of the lower limbs during walking in normal adults. Electroencephalogr Clin Neurophysiol 1989; 72:429-38.
8. Wheelwright EF, Minns RA, Law HT, Elton RA. Temporal and spatial parameters of gait in children. I: Normal control data. Dev Med Child Neurol 1993; 35(2):102-13.
9. Matsusaka N, Fujitta M, Hamamina A, Norimatsu T, Suzuki R. Relationship between right and left leg in human gait from a viewpoint of balance control. In: Winter DA, Norman R, Wells R, Hayes K, Patla A, eds. Biomechanics IX-A. Champaign: Human Kinetics Publishers, 1985; 427-30.
10. Hirokawa S. Normal gait characteristics under temporal and distance constraints. J Biomed Eng 1989; 11(6):449-56.
11. Sadeghi H, Allard P, Duhaime M. Functional gait asymmetry in able-bodied subjects. Hum Mov Sci 1997; 16:243-58.

12. Allard P, Lachance R, Aissaoui R, Duhaime M. Simultaneous bilateral 3-D able-bodied gait. Hum Mov Sci 1996; 15:327-46.
13. Sadeghi H, Allard P, Duhaime M. Contributions of Lower Limb Muscle Power in Gait of People Without Impairments. Phys Ther 2000; 80(12): 1188-1196.
14. Kadaba MP. Ramakrishnan HK. Wootten ME. Gainey J. Gorton G. Cochran GVB. Repeatability of kinematic, kinetic, and electromyographic data in normal adult gait. J Orthop Res 1989; 7(6): 849-60.
15. Olney SJ, Griffin MP, McBride ID. Multivariate examination of data from gait analysis of person with stroke. Phys Ther 1998;78:814-828.
16. Winter DA. Overall principle of lower limb support during stance phase of gait. J Biomech 1980; 13(11):923-7.
17. Hamilton HC. Regression with graphics. A second course in applied statistics. Wadsworth inc, Belmont. California, 1992.
18. Winter DA. Biomechanics and motor control of human movement. New York: Wiley-Interscience publication, John Wiley and Sons, 1990.
19. Vaughan CL. Are joint torques the Holy Grail of human gait analysis? Hum Mov Sci 1996; 15:423-43.
20. Davis BL., Vaughan CL. Phasic behavior of EMG signals during gait: use of multivariate statistics. J Electromyography and Kinesiology 1993; 3(1):51-60.
21. Sadeghi H, Prince F, Zabjek KF, Allard P. Sagittal hip muscle power during walking in old and young men. J Aging Phys Activity 2001; 9: 172-83.
22. Sadeghi H, Prince F, Sadeghi S, Labelle H. Principal component analysis of the power developed in the flexion/extension muscles of the hip in able-bodied gait. J Biomed Eng Physics. In press.
23. Apkarian J, Naumann S, Cairns B. A three-dimensional kinematic and dynamic model of the lower limb. J Biomech 1989; 22:143-55.
24. Eng JJ, Winter DA. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? J Biomech 1995; 28(6):753-58.
25. Adams J, Perry J. Gait analysis: clinical application. In: Human walking. Rose J., Gamble J eds. 2nd edition. Williams & Wilkins, Baltimore, Maryland U.S.A. 1994; 141-64.
26. Winter DA, Eng JJ, Ishac MG. A review of kinematic parameters in human walking. In: Gait analysis: Theory and application, editors. Craik and Otais. Mosby Publisher, 1995; 252-70.
27. Czerniecki JM. Foot and ankle biomechanics in walking and running. A review. Am J Phys Med & Rehabil 1988; 67:246-52.
28. Ericson MO, Nisell R, Ekholm J. Quantified electromyography of lower-limb muscles during level walking. Scandinavian J Rehabil Med 1986; 18:159-63.
29. Winter DA. Electromyography in human gait. In the biomechanics and motor control of human gait: In: Normal elderly and pathological, edited by Winter University of Waterloo Press, Waterloo, 1991; 53-73.
30. Elftman H. Forces and energy changes in the leg during walking. American Journal of Physiology 1939; 125:339-56.
31. Gray EG, Basmajian JV. Electromyography and cinematography of leg and foot during walking. Anatomical Record 1968; 161:1-15.
32. Pedotti A. A study of motor coordination and neuromuscular activities in human locomotion. Biol Cybern 1977; 26-53.
33. Winter DA. Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences. Clin Ortho & Related Res 1983; 147-154.
34. Bressler B, Berry FP. Energy and power in the leg during normal level walking. Prosthetic Devices Research Project, series 11, issue 15. Berkeley, California, Institute of Engineering Research of the University of California, 1951.
35. Inman VT. Human locomotion. J Canadian Med Ass 1966; 94(20):1047-54.
36. Perry J, Hoffer MM, Giovan P, Antonelli D, Greenberg R. Gait analysis of the triceps surae in cerebral palsy. A preoperative and postoperative clinical and electromyographic study. Am J of Bone & Joint Surg 1974; 56:511-20.
37. Simon SR, Mann RA, Hagy JL, Larsen LJ. Role of the posterior calf muscles in normal gait. Am J of Bone & Joint Surg 1978; 60:465-72.
38. Sutherland DH, Cooper L, Daniel D. The role of the ankle plantarflexors in normal walking. Am J of Bone & Joint Surg 1980; 62(A):354-63.
39. Cappozzo A, Figura F, Marchetti M. The interplay of muscular and external forces in human ambulation. J Biomech 1976; 9:35-43.
40. Dillingham TR, Lehmann JF, Price R. Effect of lower limb on body propulsion. Arch Phys Med & Rehabil 1992; 73:647-51.
41. Meinders M, Gitter A, Czerniecki JM. The role of ankle plantar flexor muscle work during walking. Scandinavian J Rehabil Med 1998; 30:39-46.
42. Sutherland DH, Kaufman KR, Mottoza JR. Kinematics of normal human walking. In Rose J, Gamble JG eds. Human walking: second edition. Williams & Wilkins. Baltimore. 1994: 25-44.
43. Winter DA. Kinematic and kinetic patters in human gait: variability and compensating effects. Hum Mov Sci 1984; 3: 51-76.

44. Gundersen LA, Valle DR, Barr AE, Danoff JV, Stanhope SJ, Snyder-Mackler L. Bilateral analysis of the knee and ankle during gait: an examination of the relationship between lateral dominance and symmetry. Phys Ther 1989; 69(8):640-50.
45. Gray EG, Basmajian JV. Electromyography and cinematography of leg and foot during walking. Anatomical Record 1968; 161(1):1-15.
46. Winter DA, Patla AE, Frank JS. Assessment of balance control in humans. Med Progress through Tech 1990;16: 31-51.
47. Kepple TM, Siegel KL, Stanhope SJ. Relative contributions of the lower extremity joint moments to forward progression and support during gait. Gait & Posture 1997; 6:1-8.
48. Perry J. Gait analysis: normal and pathological function. Thorofare (NJ): Slack; 1992.
49. Winter DA, Eng P. Kinetics: Our window into the goals and strategies of the central nervous system. Behavior Brain Res 1995; 67:111-20.
50. Judge JO, Davis RB III, Ounpuu S: Step length reduction in advanced age: the role of ankle and hip kineticts. J Gerontal Med Sci 1996; 51A: M303-M12.