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IJIE

Journal homepage: http://penerbit.uthm.edu.my/ojs/index.php/ijie
ISSN : 2229-838X e-ISSN : 2600-7916

The International Journal of Integrated Engineering

Instantaneous Segmental Energy Symmetry Index as Gait Compensation Indicator in Asymmetrical Walking

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DOI: https://doi.org/10.30880/ijie.2019.11.03.015
Received 5 March 2019; Accepted 4 July 2019; Available online 3 September 2019

Abstract:
Purpose: Human body constantly adapts to optimise the energy expenditure. A better understanding of the mechanical energetic costs in lower extremities helps identify the compensatory mechanism adopted in asymmetrical gait. This paper proposes the use of instantaneous segmental energy and normalised symmetry index (SI_{norm}) to examine asymmetrical gait. This approach can provide better overview of gait quality allowing identification of change in segmental energy during different gait phases and contribution of each segment in compensating abnormal walking.
Method: An experimental study was carried out to validate this method. Twenty healthy subjects were recruited. Asymmetrical gait was simulated by restricting knee motion during walking using a knee brace. Mechanical energy was determined for each segment of the left and right limbs. Normalised Symmetry Index (SI_{norm}) was then calculated to examine bilateral differences in segmental energy during stance phase and swing phase. Statistical analysis using ANOVA and Tukey-Kramer multiple comparison test to identify asymmetry of the segmental energy (p-value < 0.05).
Result: Significant asymmetry of segmental energy occurred during swing phase. Greater asymmetry was observed in kinetic energy than in potential energy. The affected limb segments produced lower kinetic energy than the normal limb. At asymmetrical state, potential energy of the affected limb's foot and thigh were lower than that of the normal segments while the inverse was true for thigh segment.
Conclusion: These results suggested that in asymmetrical gait, a form of compensatory mechanism is adopted to walk. This can be observed in the change of instantaneous segmental energy during walking.

Keywords: gait symmetry, energetic cost, normalized symmetry index

1. Introduction

Understanding the energy expenditure of human locomotion has been an area of great interest in gait analysis [1]. Energy expenditure can be determined either by quantifying the metabolic or mechanical energetic cost. The first one is the most direct approach. It is derived from the prediction of metabolic function of the body through proxies such as the body demand for adenosine triphosphate (ATP) during an activity, which is reflected in the change in volumetric rate of oxygen consumption. The rate of oxygen consumption signifies the rate of cellular respiration, which is directly proportional to the intensity of the activity [1]. However, this method only measures how much energy is expended by

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the whole body as a single system without differentiating the source of energy expenditure. The second method can overcome this limitation. Mechanical energetic cost is derived from the interaction between potential and kinetic energies during walking. It can measure the instantaneous energy profile of human lower extremity and their changes in gait. Several studies reported the relationship between the mechanical and metabolic energies [2]. It was found that metabolic energetic cost increases when there is an increase or decrease in stride rate from the preferred rate of the subjects for different reasons [3]. This change is attributed to an increase in the mechanical work done by the lower extremity to propel the body forward.

Human gait is defined as a cyclical movement pattern of the limbs during locomotion. The most important and commonly studied gaits are walking and running gaits. Walking gait is divided into two main phases of swing phase and stance phase. These phases are defined using the toe off and heel strike events [4]. They are then further divided into several key events and periods as shown in Figure 1.

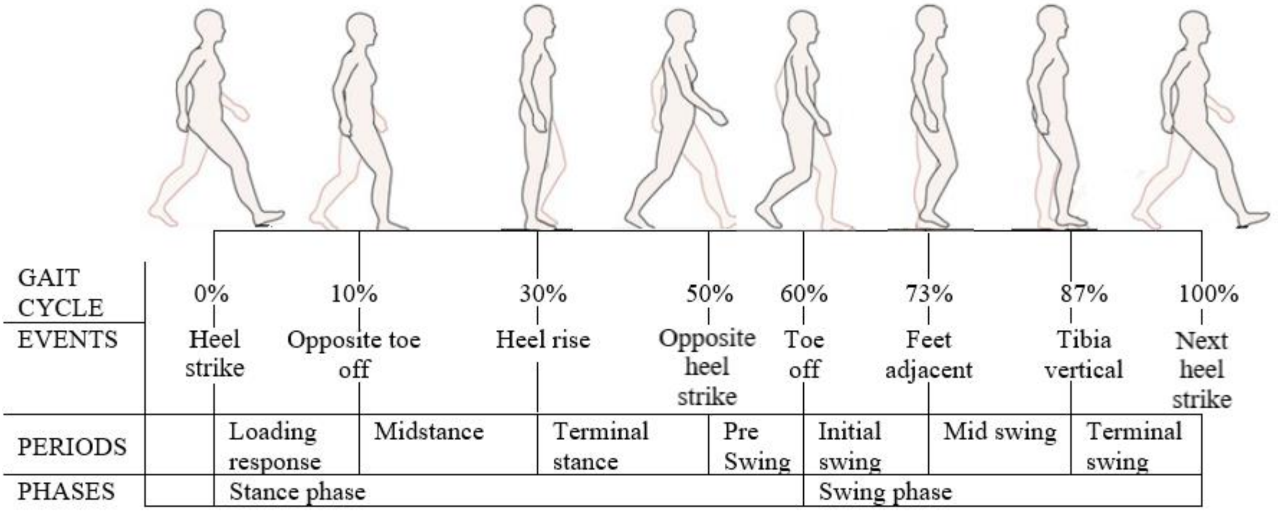

Fig. 1 - Key events and phases in a complete gait cycle. Modified from [5].

Gait of a healthy individual is fairly symmetrical with small deviations. Patients with musculoskeletal disorders (e.g. lower limb joint immobilisation due to application of plaster cast [6], unilateral osteoarthritis [7] and patients with neurological disorders e.g. stroke [8, 9]) generally exhibit asymmetrical gait. Significant bilateral differences between left and right limbs during locomotion can be observed in these patients. Several indices have been proposed to define gait asymmetry, such as symmetry ratio [9, 10] and symmetry index [9, 11, 12], in patients with stroke [10], cerebral palsy [13] and amputation [14]. Despite their wide adoption in clinical and rehabilitation settings, these indices are subjected to artificial inflation, especially when the gait parameters are close to zero. Normalised Symmetry Index (SI_{norm}) was reported to be able to overcome this limitation and assess gait parameters with continuous waveform without being subjected to this artifact [12].

Change in the gait can affect the mechanical energetic cost of walking. In normal gait, energy is continuously optimized by the selection of a nominal gait to minimise the energy expenditure [6, 15]. As such, individuals tend to have their own preferred gait parameters. One simple and quantifiable mechanical energetic cost of the lower extremity is the instantaneous kinetic and potential energies of the limb segments [16]. When asymmetrical gait is present, the normal gait parameters and subsequently the energetic cost will be disrupted. It also induces additional stress to the limbs.

Although the underlying pathological and neurological conditions that lead to asymmetrical gait has been studied, the bilateral energy distribution between the two lower limbs remain unclear. Moreover, despite extensive studies on gait asymmetry, most of them are limited to gait spatial and temporal parameters such as gait phase time [9, 17], step length, and ground reaction force profile [9]. None of them involves mechanical energy of the lower extremity. This study is a continuation of our previous works reported in [18] and [19]. In [18], we presented the kinematic and kinetic aspects of the asymmetrical gait, while in [19], we determined the bilateral difference between left limb and right limb in asymmetrical gait and examined the change in mechanical energy of each segment (foot, shank and thigh). From these studies, we hypothesize that there will be significant difference between segmental energies of the left and right limbs during certain period of the gait cycle in asymmetrical gait. Therefore, statistical analysis was performed here to further examine the significance of the asymmetrical gait throughout the gait cycle and to determine the period in which the energy interaction between left and right limbs might have occurred. It is expected that this study can help researchers to gain better understanding on asymmetrical gait and devise better treatment and rehabilitation plan and engineers to design better orthoses or prostheses with improved gait symmetry, greater stability and less stress on the patients' lower extremity joints.

2. Methods

Ten male and ten female healthy subjects (Age: 22.4 ± 1.96 years old; Height: 165.7 ± 10.16 cm; Weight: 61.16 ± 14.06 kg) with no known history of lower extremity injuries were recruited. This study was approved by Monash University Research Ethics Committee. The gait data were collected using six Oqus® infrared camera motion capture system(Qualisys Inc.) and two forceplates (Bertec Co.) with sampling frequency of 100 Hz.

Thirty four reflective passive markers were placed on the subjects' lower extremities (from waist to toe) according to a modified Cleveland Clinical markers system (Figure 2) before any trials. The subject was then requested to walk as naturally as possible on a flat 10m platform embedded with two force plates at a comfortable walking speed of their own choice while barefooted under 3 different conditions: without brace on (Normal), with orthopaedic knee brace (Trom Advance Knee Brace, DonJoy) one limb at a time on the left (AbLeft) and right lower limbs (AbRight). The orthopaedic knee brace was set to a fixed degree to restrict knee movement and to simulate stiff knee condition. For each condition, 10 walking trials and one static trial were taken. The subjects were allowed to rest between each conditions' trials.

Gait data were imported into Visual 3D (C-Motion Inc.) to derive relevant gait parameters. Gait events such as heel-strike and toe-off were determined using method described in [20]. They were also used to partition the data into two main phases: stance phase and swing phase. The kinetic and potential energies of foot, shank and thigh were calculated as described below.

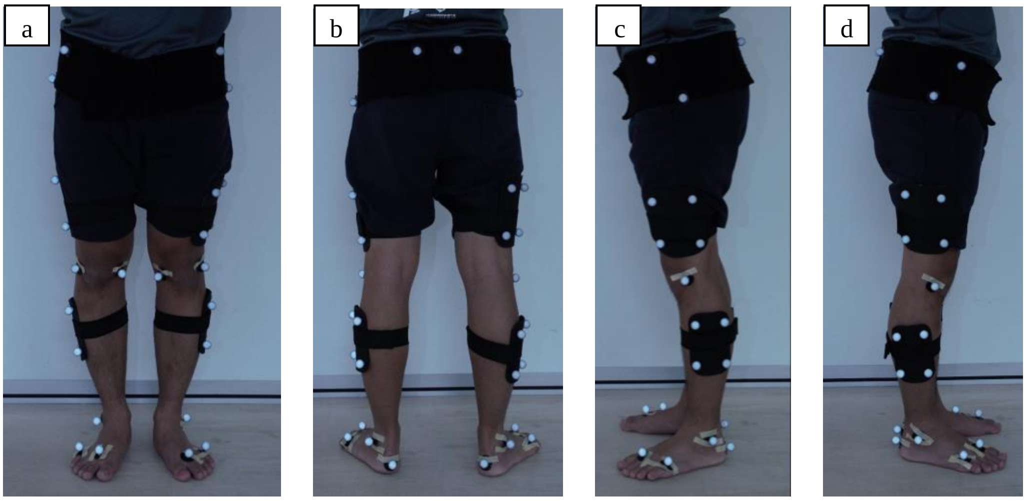

Fig. 2 - Markers positions on the lower extremities of a subject with no knee brace on (Normal condition) (a) in front, (b) back, (c and d) side views

The kinetic energy (T) comprises of translational (T_T) and rotational kinetic energies (T_R), as defined in (1) and (2).

$$T_T = \frac{1}{2} m v^2 \tag{1}$$

Where: $v = v_x^2 + v_x^2 + v_x^2$, v_x = Velocity in x-axis direction, v_y = Velocity in y-axis direction and v_z = Velocity in z-axis direction.

$$T_R = \frac{1}{2}(I_{xx}\omega_x^2 + I_{yy}\omega_y^2 + I_{zz}\omega_z^2) \tag{2}$$

Where: I_{xx} = Inertia of moment about x-axis, I_{yy} = Inertia of moment about y-axis , I_{zz} = Inertia of moment about z-axis, ω_x = angular velocity in x-axis, ω_y = angular velocity in y-axis and ω_z = angular velocity in z-axis. The potential energy (U) of each body segment is defined in (3).

$$U = mgh \tag{3}$$

Where: m = segment mass, g = gravity (9.81 m/s) and h = the position of the segment COM (Center of Mass) in the vertical direction (COM_z).

A script written in MATLAB® R2017a (MathWorks, Inc) was used to further process the gait data. Translational and rotational energies were normalised by the segment mass while potential energy was normalized by the segment mass and the segment height measured from the ground.

The normalized kinetic energy (T_{norm}) was then obtained using (4) where T_{Tnorm} and T_{Rnorm} are the normalized translational energy and rotational energy respectively.

$$T_{norm} = T_{Tnorm} + T_{Rnorm} \tag{4}$$

The gait cycle was then divided and segmented into two phases: swing phase and stance phase. Each phase was interpolated to 100 data points. The means of each data point of the segmental energies were calculated and normalised using min-max normalisation as described in (5) with the corresponding left limb segment's energy being the reference.

$$Normalised\_Energy, E_{norm} = \frac{E_n - E_{\min}}{E_{\max} - E_{\min}} + 1 \tag{5}$$

Where: E_n = Energy at n point ($n = 1, 2, 3 \ldots 100$), E_{min} = Minimum energy of left limb segment of corresponding gait condition, E_{max} = Maximum energy of left limb segment of corresponding gait condition.

Normalised symmetry index (SI_{norm}) of each data points of each subject was then calculated using (6).

$$SI_{norm} = \frac{E_{Lnorm}(t) - E_{Rnorm}(t)}{0.5(E_{Lnorm}(t) + E_{Rnorm}(t))} \tag{6}$$

Where: $E_{Lnorm}(t)$ = Normalised energy of left segment, $E_{Rnorm}(t)$ = Normalised energy of right segment, t = Time (percentage of gait phase, $t = 1, 2, 3 \ldots 100$).

To further examine significant differences of the three gait conditions, ANOVA was used. The significance level (alpha value) was set at 0.05. Tukey-Kramer multiple comparison test was performed for the data points with $p < 0.05$ to further examine the pairwise differences at different walking conditions.

3. Results

The normalised kinetic and potential energies of one subject are presented in Figure 3. The change in kinetic energy is more apparent during the swing phase rather than the stance phase (Figure 3(a) and (b)). The kinetic energy reaches its maximum during the early swing phase in Normal condition (approximately 12.5 J/kg) and in AbLeft (approximately 11 J/kg). No such peak is observed in AbRight condition – The kinetic energy remains relatively constant throughout the swing phase. On the other hand, during stance phase, the differences among the three walking conditions can be observed in the early and end of stance phase.

Greater shank potential energy was recorded during AbRight than during Normal and AbLeft throughout the whole gait cycle. For potential energy, the braced leg of AbRight recorded higher potential energy in the shank compared to unbraced leg of AbLeft and Normal throughout the whole gait cycle. A peak occurred in Normal during the initial swing phase to about 38 J/kgm whereas for AbLeft, the peak occurred earlier in the swing phase and was lower (35 J/kgm).

Figure 4 and 5 depict the graphical comparison of the mean of the SI_{norm} of segmental kinetic and potential energies of the foot, shank and thigh at three different walking conditions. It can be observed that the mean of the SI_{norm} in normal walking condition are near zero throughout the stance and swing phases. These results indicate that there are minor differences between right and left segments, and thus implying that the segmental energy is symmetrical. They match the convention that the gait of healthy individual is symmetrical with minor deviations. It also shows that both limbs are contributing an equal amount of mechanical energy needed for walking and neither one of the limbs is stressed more than the other.

From the means of the SI_{norm} for AbLeft and AbRight, it was noted that there were some asymmetry of segmental energies, with the kinetic segmental energies showing greater asymmetry than the potential segmental energies (except for the foot segment) especially during swing phase. The waveforms for the means of the SI_{norm} for AbLeft and AbRight were noted to be almost mirrored around the x-axis. Due to the equations used, positive SI_{norm} means that the energy of the left limb segment is greater than that of the right limb segment.

Table 1 presents the summary of time duration in the gait cycle where null hypothesis was rejected in ANOVA ($p < 0.05$) for the SI_{norm} of kinetic and potential energy for the segments of interest. ANOVA only indicates significant differences among the three conditions and does not describe whether there is significant asymmetry in the abnormal conditions. Since the mirrored AbRight and AbLeft are vastly different, it causes a false positive to be detected in ANOVA as the significant differences from ANOVA might only be caused by differences between the AbRight and AbLeft conditions. Thus, pairwise Tukey-Kramer multiple comparison test was conducted for the period when $p < 0.05$. Table 2 and 3 present the summary of time duration and gait events where null hypothesis was rejected in pairwise Tukey-Kramer multiple comparison test between the SI_{norm} of kinetic and potential energies respectively for the segments of interest for abnormal conditions (AbRight and AbLeft) against the Normal condition.

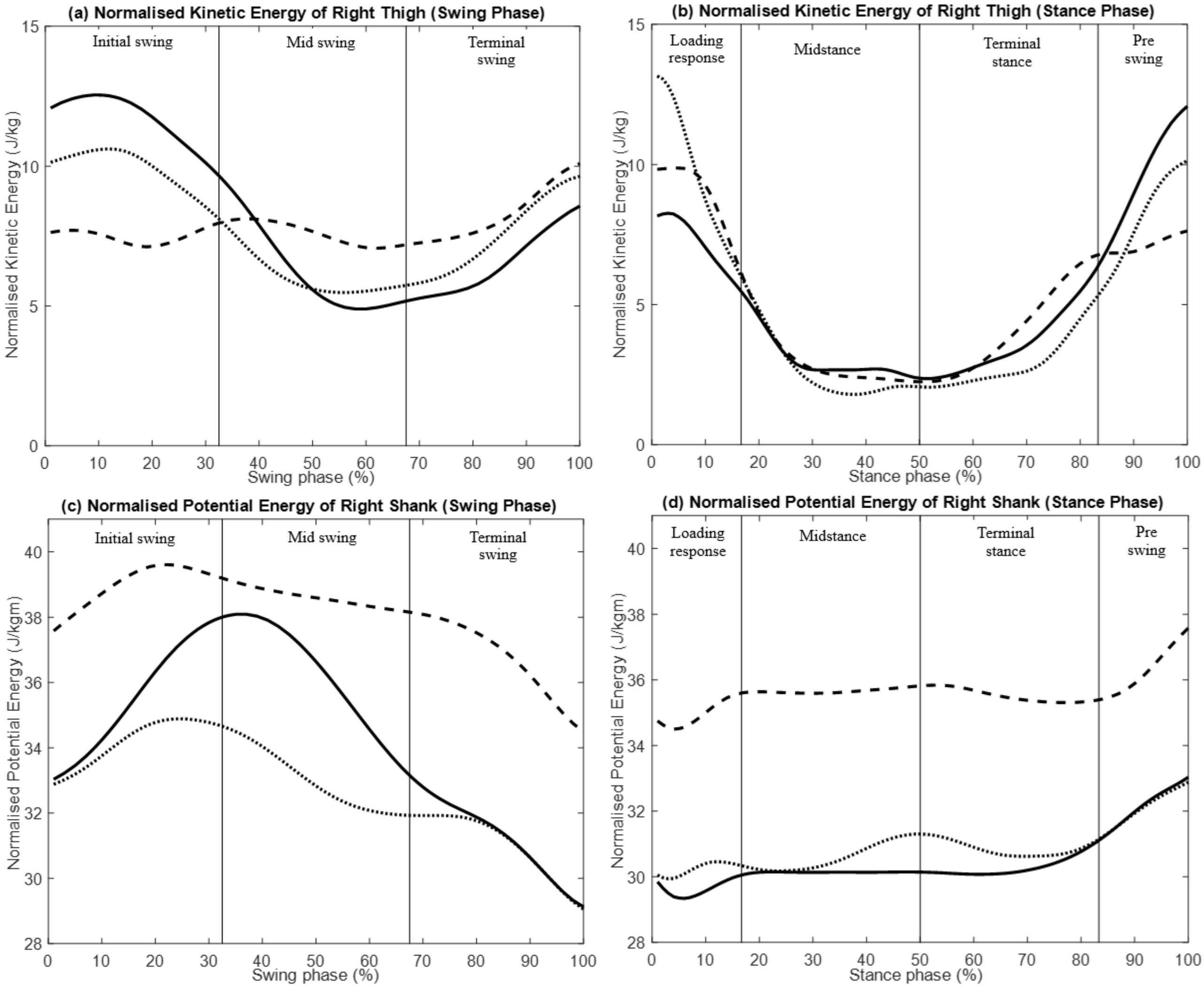

Fig. 3 - Normalised (a, b) kinetic energy of the right thigh and (c, d) potential energy of the right shank of one subject against gait phase (%). (Solid line – Normal; Dashed line – AbRight; Dotted line – AbLeft)

Table 1 - Summary of Kinetic and Potential Energy SI_{norm} ANOVA.

Energy	**Segment**	**Time duration where $p < 0.05$**	
		Swing phase (%)	**Stance phase (%)**
Kinetic	Foot	1 – 13	1 – 2
		21 – 83	87 – 100
		91 – 100	
	Shank	1 – 49	7 – 18
		75 – 100	30 – 69
			84 – 100
	Thigh	1 – 32	4 – 37
		37 – 96	47 – 60
			76 – 100
Potential	Foot	1 – 54	90 – 100
		61 – 75	
	Shank	6 – 60	-
	Thigh	1 – 92	57 – 80
			94 – 100

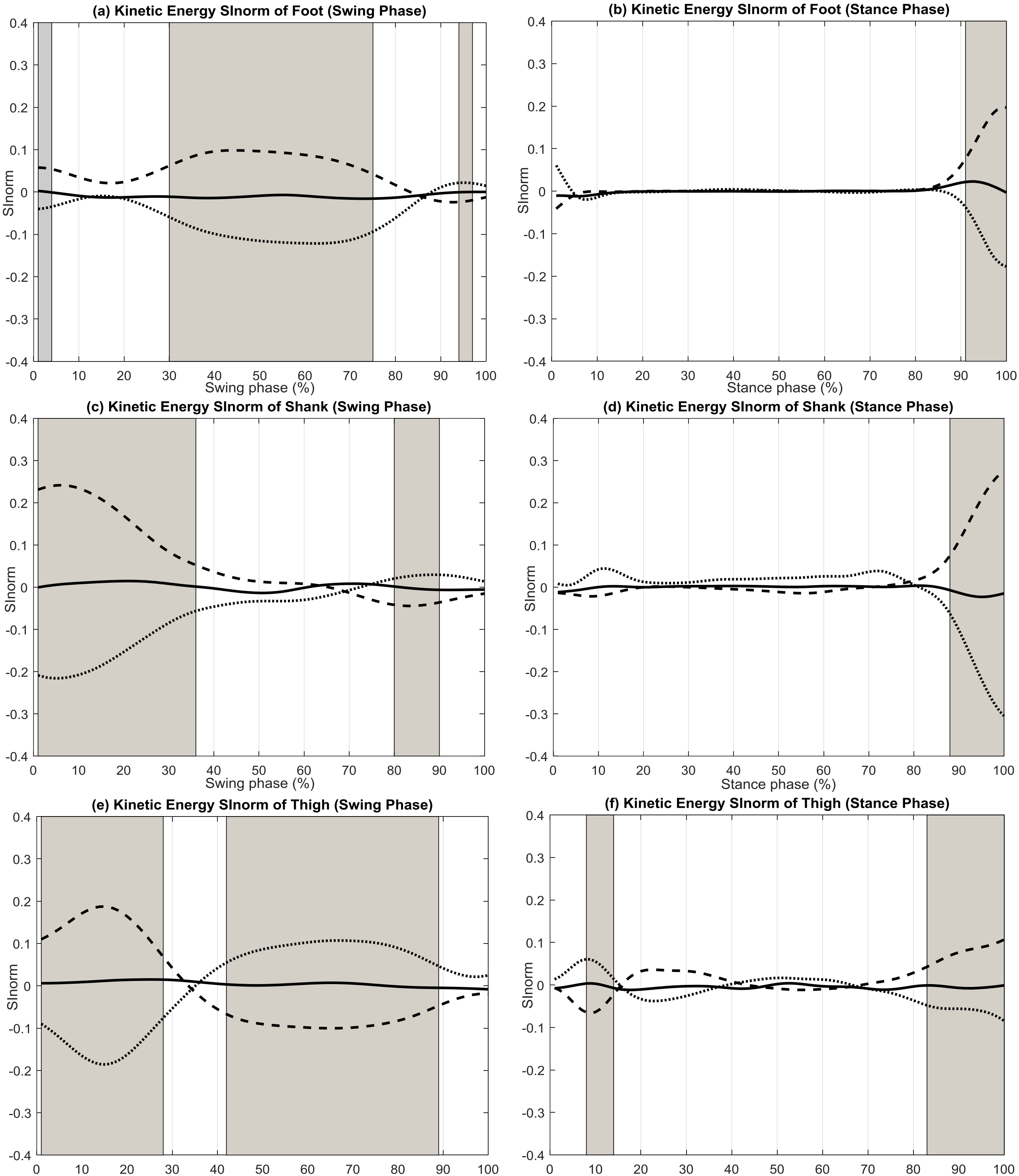

Fig. 4 – Kinetic energy SI_{norm} against gait phase (%) of the (a, b) foot, (c, d) shank and (e, f) thigh segments. Shaded area represents the % phase where significant asymmetry of kinetic energy occurred. (Solid line – Normal; Dashed line – AbRight; Dotted line – AbLeft)

Table 2 - Summary of Kinetic Energy SI_{norm} Tukey-Kramer multiple comparison test pairwise comparison of abnormal gaits (AbLeft and AbRight) with normal gait (Normal).

Pairwise comparison	Segment	Time duration where $p < 0.05$		
		Swing phase (%)	Stance phase (%)	Gait event
AbLeft with Normal	Foot	1 – 4 30 – 80 94 – 99	1 91 – 100	Heel strike, pre-swing, toe off, mid swing, terminal swing
	Shank	1 – 37 86 – 93	8 – 17 33 – 68 88 – 100	Loading response, mid stance, terminal stance, pre swing, toe off, initial swing, terminal swing
	Thigh	1 – 32 42 – 91	7 – 9 24 – 26 82 – 100	Loading response, mid stance, pre swing, toe off, initial swing, mid stance, terminal swing
AbRight with Normal	Foot	1 – 12 24 – 75 93 – 97	91 – 100	Pre swing, toe off, initial swing, mid swing, terminal swing
	Shank	1 – 36 76 – 89	45 – 57 86 – 100	Mid stance, terminal stance, pre swing, toe off, initial swing, terminal swing
	Thigh	1 – 28 38 – 89	7 – 11 19 – 36 83 – 100	Loading response, mid stance, pre swing, toe off, initial swing, mid stance, terminal swing

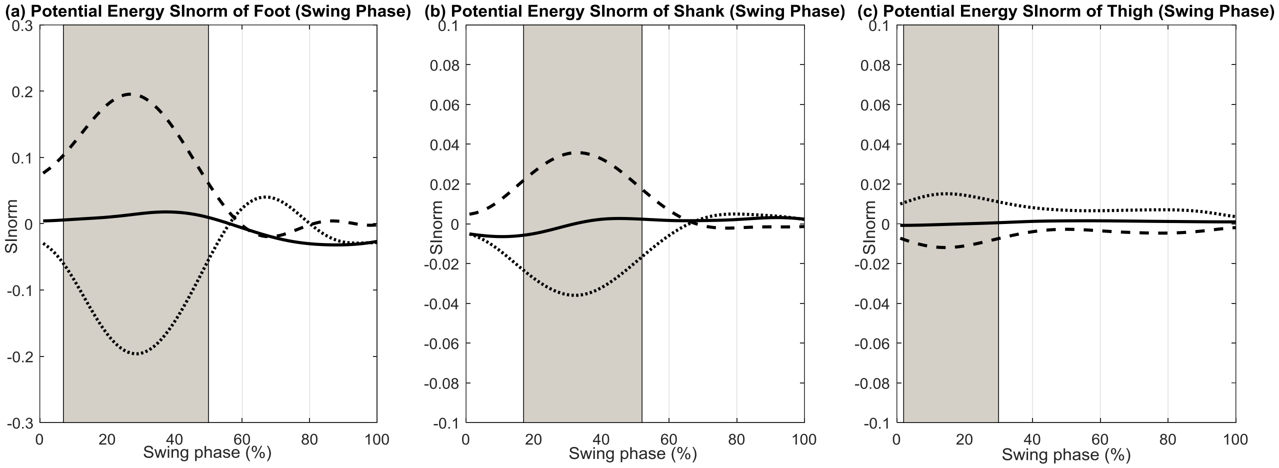

Fig. 5 - Potential energy SI_{norm} against swing phase (%) of the (a) foot, (b) shank and (c) thigh segments. Dashed lines represents the standard deviations of the data. Shaded area represents the % phase where significant asymmetry of potential energy occurred. (Solid line – Normal; Dashed line – AbRight; Dotted line – AbLeft)

Table 3 - Summary of Potential Energy SI_{norm} Tukey-Kramer multiple comparison test pairwise comparison of abnormal gaits (AbLeft and AbRight) with normal gait (Normal).

Pairwise comparison	Segment	Time duration where $p < 0.05$		
		Swing phase (%)	Stance phase (%)	Gait event
AbLeft with Normal	Foot	7 – 51	-	Initial swing, mid swing
	Shank	17 – 56	-	Initial swing, mid swing
	Thigh	1 – 37	96 – 100	Toe off, initial swing
AbRight with Normal	Foot	1 – 50	98 – 100	Toe off, initial swing, mid swing
	Shank	7 – 52	-	Initial swing, mid swing
	Thigh	2 – 30	100	Toe off, initial swing

For kinetic energy of the foot segment, energy asymmetry occurred during pre-swing, toe off and during most of the middle part of the swing phase of the gait cycle. At the end of stance phase, when the significant asymmetry of SI_{norm} between Normal with AbRight and AbLeft occurred, the SI_{norm} peak and through for AbRight and AbLeft were 0.2 and -0.2 respectively. While for shank, prominent asymmetry occurred starting from pre-swing until the end of the initial swing phase. As in the foot segment, the highest discrepancy of SI_{norm} between Normal with AbRight (highest peak ≈ 0.25) and AbLeft (lowest trough ≈ -0.25) also occurred during end of stance phase and start of swing phase. In thigh segment, only during the initial stance phase that asymmetry was not significant. The greatest significant asymmetry recorded was when the AbRight peak and AbLeft trough reached over 0.2 and -0.2 respectively during the initial swing phase. The kinetic energy of the segments of the abnormal limb were noted to be less than that of the normal limb's during phases when there was significant asymmetry, except during the mid to terminal swing phase for the thigh segment energy where the opposite is true (Figure 4).

The asymmetry of potential energy was not as significant as compared with that of kinetic energy because change in COM_Z was not large. For foot and shank segmental potential energy, significant energy asymmetry occurred from the initial swing to mid swing in the gait cycle. In the thigh segment, the asymmetry of potential energy SI_{norm} occurred during toe off to the end of the initial swing phase. The potential energy SI_{norm} AbRight peak (0.2) and AbLeft trough (-0.2) of for foot segment were greater than that of shank (AbRight peak ≈ 0.04 and AbLeft trough ≈ -0.04) and thigh (AbLeft peak ≈ 0.02 and AbRight trough ≈ -0.02) segment. During durations of asymmetry, potential energy of the foot and shank segments of the affected limb were lower than that of the normal limb. While for thigh segment the inverse was true (Figure 5).

4. Discussion

This study shows that identifying asymmetrical gait using the mechanical energy for SI_{norm} is plausible. It allows the asymmetrical behavior of the instantaneous energy of the limb segments to be easily identified. It also offers a different way of interpreting asymmetrical gait, instead of the usual temporal and kinematic gait parameters. Instantaneous energy gives a better understanding on the effects of asymmetrical gait on the energetic cost of the lower limbs.

The mirroring of the waveforms for the means of SI_{norm} for AbLeft and AbRight implies that the changes in the segmental energies caused by the abnormality were similar regardless of whether the abnormality occurs on the left or right limb. This was due to the fixed reference side (left limb segment) used in the calculations. Selecting one side of the limb to be the reference segment is reasonable because in an uncontrolled environment, the affected and the non-affected limb may be unknown.

The lower kinetic energy of the abnormal limb implies that the abnormal limb moved at a lower velocity than the normal limb during their respective swing phases. Therefore, to maintain the same walking speed as in normal gait, the normal limb moved at higher velocity, which in turn increases the energetic cost of walking. Despite so, the walking speed during abnormal conditions were still lower than the normal condition. This is consistent with past findings that recorded an overall decrease in walking speed in stroke patients [21], patients with stiff knee gait (SKG) [8] and subjects with simulated SKG [22].

Even though the normal limb's thigh segment had higher kinetic energy than that of the affected limb's at the beginning of swing phase, by the second half of the swing phase to the early stance phase (Figure 4(c)), its kinetic energy was lower than that of the affected limb's. This means that the normal limb was able to brake much more efficiently than the affected limb to slow down the movement before heel strike. Therefore, from this observation, it can be said that the efficiency of the braking mechanism of the affected limb has been reduced. For stroke patients, there is absence of braking mechanism at the knee. The braking mechanism at the knee is produced by thigh muscles. In a patient, the function of the thigh muscles will be impaired [23]. Based on the findings from literature, this trend should have been seen in the shank segment. Instead, for this experiment, the impaired braking mechanism was observed in the thigh segment while it was not present in the shank segment. One of the possible reasons could be the participants were healthy with normal functioning thigh muscles and thus the segment where the braking mechanism was affected differed from that of a stroke patient's.

A greater potential energy meant that the COM_Z of the segment and thus the limb segment was lifted higher above the ground. This implied that the foot and shank segments of the affected limb were lower than that of the normal limb during the same period in the gait cycle respectively while the inverse was true for the thigh segment. This is consistent with past studies that found that in SKG, due to the knee flexion being limited and thus the affected limb's knee not being able to flex to lift the shank and foot (accounting for the lower COM_Z). To avoid foot drag, compensation in the form of hiking of the pelvis and increased hip circumduction on the affected side (higher COM_Z of the affected limb's thigh segment) was observed [22]. There will also be contralateral vaulting gait which is characterised by the artificial lengthening of the contralateral (normal in this case) limb by raising the heel and keeping the knee joint locked in full extension to compensate for the increased length of the swinging limb caused by the rigid and extended knee joint [22].

This vaulting would explain why the potential energy of shank and foot segments of the normal limb in AbRight and AbLeft trials were higher than that of the abnormal limb.

This study was limited to the simulation of SKG by limiting knee flexion in healthy subjects, patients with the SKG may have different walking patterns. The patients feel pain, tire easily due to increased energetic cost of locomotion [22] and walk at a slower pace [8, 21]. Thus, further studies on these patients are needed. Nonetheless, the results obtained in this study suggested that SKG was sufficiently simulated in the healthy subjects such that the subjects displayed several traits seen in real patients, including reduced velocity and energy (Figure 4) of the affected limb which in turn caused a decrease in the subjects' walking speed during the AbRight and AbLeft trials compared to the normal trials. The subjects also displayed some forms of compensation seen in real patients. The advantage of using healthy subjects is that the normal gait trials will provide a good baseline to compare with that of the simulated abnormal gait trials since this will eliminate inter-subject deviation. Another limitation would be that instantaneous energy does not really give a clear picture of the transfer and generation of energy and their interactions between the segments and joints in gait. To better understand the energy flow between the segments, a 3D musculoskeletal model of the lower extremity is required and will be explored in our future study.

5. Conclusion

Asymmetrical gait was successfully simulated in this study. It shows that it is possible to identify asymmetrical gait using the lower limb segmental mechanical energy and SI_{norm}. In a simulated SKG, significant asymmetry of segmental energy occurs during the swing phase of the gait cycle. Greater asymmetry was observed in the kinetic energy than in the potential energy. To compensate for the bilateral differences between the limbs, the affected shank and foot produce lower kinetic energy and potential energy than the unaffected limb while the affected thigh generates lower kinetic energy and higher potential energy than the unaffected thigh.

Acknowledgement

This research is supported by Malaysia Ministry of Higher Education (Project Ref. No. FRGS/1/2016/TK03/MUSM/02/1).

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