Chapter

A Similarity Index for Balance Assessment between Older Adults with and without Balance Deficits

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Abstract

Falls in older adults can cause disabling health even though falls are largely preventable. A combination of fall risk factors can be modified or predicted to minimize devastating complications. However, clinical balance assessment tools often have contradictory results since fall risks are individualized and multifactorial. The assessment tools are often practically limited to detecting sensitive changes between older adults with and without balance deficits. Recently, a similarity index (SI) has been developed to predict fall risks based on kinematic data during gait. The combined limb motions to those of a prototype derived from healthy individuals in the gait cycle might be differentiated from individuals with gait pathologies. The analyzed calculations result in response vectors that would be compared to controls-derived prototype response vectors. Furthermore, the normalized SI, based on the vector representing the data distribution, could be generated from the enhanced (dis)similarities dataset of subjects following an intervention (prototype response vectors). These quantified indices for compensatory patterns provide a further understanding of optimal injury prevention and specific rehabilitation strategies for older adults with balance deficits. This chapter will propose a novel sensitive measure, the SI, for older adults with orthopedic and neurologic dysfunction compared with control subjects.

Keywords: kinematic, similarity index, balance deficits, pain, older adults, gait cycle, motions

1. Introduction

Balance problems in older adults are of major concern as functional declines of the somatosensory system occur in aging populations, potentially contributing to postural instability [1, 2]. These problems provide foundational knowledge of balance performance and the importance of using a reliable and valid sensory testing protocol for older adults. However, valid balance measurements are burdensome and costly, especially for older adults with balance deficits who are characterized by greater comorbidity compared to healthy adults [3]. As a result, it would be critical to evaluate the characteristics of falls and clarify the advantages of predicting the occurrence of falls in older adults with balance deficits.

Previous studies utilized clinical measure tools, such as the Berg Balance Scale (BBS), the Dynamic Gait Index (DGI), as well as other advanced balance measures [4–6]. However, the small sample size in their studies limits the validity of the results to generalize measurement outcomes. A sensitive balance detection tool with a larger sample size can be compared to examine balance deficits. It would be valuable to utilize a valid tool based on both scientific understanding and clinical applications of balance mechanisms in older adults. Previous research reports did not necessarily prove the sensitive outcome assessment by detailing how the feasibility of those measurement tools will achieve fall assessment and prevention. However, our studies attempted to comprehensively evaluate outcomes by introducing biomechanical research and clinical applications on biomechanics and neuromuscular control during functional activities [7–9].

A similarity index (SI) tool provides a detailed picture of different limb and trunk muscle activations by electromyography (EMG) and kinematic stability during gait. Most older adults with balance deficits present with impaired postural control [10–12] and motor coordination, including the inability to initiate, continue, or terminate activation of multiple muscle groups in a task- and environment-specific manner [13, 14]. However, the existing balance assessment tools often lack clinical evidence and demonstrate conflicting outcomes. It is critical to compare age- and gender-matched groups with balance deficits using clinical measurement tools. These tests evaluate dynamic stability in older adults with balance deficits related to somatosensory impairments in musculoskeletal and neurologic symptoms.

2. Gait assessment in individuals with musculoskeletal dysfunction

Decreased physical activity levels result in a concomitant decline in musculoskeletal function. Adequate muscular strength is fundamental to preserving functional mobility from asymmetrical limb motions during gait, which has been an increasingly important theme of research in recent years [15–17]. These studies predicted the risk of falls from the gait measures, and there is an increased risk of multiple falls in older adults with poor gait. Specific measures of gait and gait variability from musculoskeletal dysfunction confer the balance deficits and could be amenable to reducing the risk of falls.

Neck dysfunction/pain is the fourth leading cause of disability and the most common musculoskeletal disorder in primary care [18, 19]. Several studies reported that individuals with neck pain have impairments of balance and head-neck coordination as well as sensorimotor deficits including poorer proprioception [20–22]. However, there is a lack of understanding of gait parameters and the kinematic SI on the limbs during gait. The similarity of gait patterns and compensated limb motions in individuals with neck pain compared to healthy controls was not carefully investigated. If the similarity of the combined limb motions is lessened in individuals with neck pain compared with healthy controls, then those kinematic changes might be detected in the specific phases of the gait cycle. The magnitude of motions and the degree of similarity between groups may provide clinical insights to enhance a comprehensive understanding of the functional consequences of gait deviations.

A recent study indicated that SI was a useful measure to differentiate similarities between groups at specific phases during gait [23]. The kinematic SI during gait was investigated to compare the ratio between the vector representing the distribution of the motions in individuals with neck pain and that representing the average distribution in individuals without neck pain. These SI values of the control group were significantly higher than the neck pain group during gait, especially in the midstance and swing phases. The SI could have the potential to highlight differences in the neck pain group during gait more than common parameters, such as cadence, speed, stride length, and step width. Ultimately, an objective measure, such as the SI, may enhance gait evaluations and help to determine the similarity of combined kinematic changes during gait.

The SI measure could provide an objective tool for the overall kinematic variations and similarities that occur during gait between groups as well. For example, an altered gait pattern was evident in individuals with neck pain who demonstrated a slower and more asymmetrical gait [24]. Therefore, the similarities of the combined upper and lower limbs might clarify the difference in gait patterns between individuals with and without neck pain. It is valuable to identify the specific phases of gait and to compare the similarity between individuals with and without neck pain. The normalized SI measures were from three-dimensional (3D) motions and provide quantitative evidence at 5% intervals of the entire gait cycle. The collected data may guide design to improve gait function.

2.1 Methodology of the kinematic similarity index

Spatiotemporal and kinematic parameters were calculated for each gait cycle using OrthoTrac 5.2 software (Motion Analysis Corporation, Santa Rosa, CA, USA). The spatiotemporal parameters included cadence, gait speed, stride length, and step width. Three-dimensional upper and lower limb kinematic waveforms were time normalized to 100 points, comprising the gait cycle from 0 to 100% with 1% increments [25, 26].

The kinematic SI computation is a numerical expression of the motion similarity in the response vectors (RV) between participants with and without neck pain (NP). The SI was computed as the normalized cosine angle (θ), where θ is the angle between two vectors, which were prototype and response vectors. The prototype response vector (PRV) represented the distribution of activity generated by participants without neck pain (NP), and the response vector (RV) represented the distribution of the participants with neck pain (NP). The RV was a series of elements with angles of each joint at a specific time point during gait for each subject. The PRV was an average of response vectors from the control participants. The mathematical equation for computing SI, the cosine of the angle between two vectors, is indicated in Equation (1).

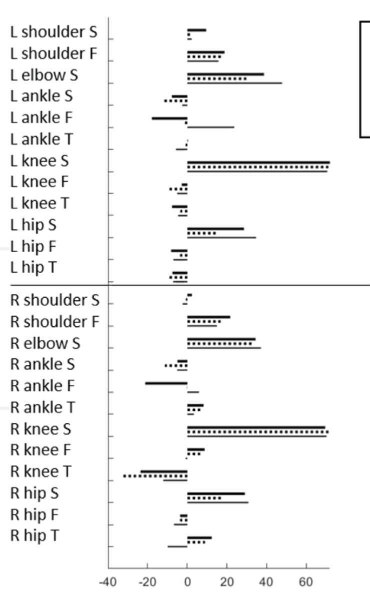
The kinematic data included the bilateral shoulders, elbows, hips, knees, and ankles. However, the elbow did not possess frontal or transverse motions, and the shoulder possessed negligible transverse motion during gait. To quantify the gait patterns, the SI was computed from kinematic data using Equation (1), which is the summation of the corresponding elements (numerator), divided by the magnitude of both vectors (denominator).

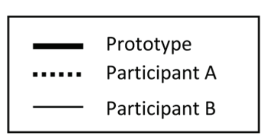
$$SI_x = \frac{\sum_i \left(PRV_{x_i} . RV_{x_i} \right)}{|PRV_x||RV_x|} \tag{1}$$

PRVx= [PRVx_1, PRVx_1, ... PRVx_n] at x% during gait. PRVx_1: averaged value of L-Shoulder flexion angles (S) from all control participants. PRVx_2: averaged value of L-Shoulder abduction angles (F) from all control participants. PRVx_n: averaged value of n'th channel from all control participants. RVx= [RVx_1, RVx_2, ... RVx_n] at x% during gait. RVx_1: L-Shoulder flexion angles (S) from a participant with NP. RVx_2: L-Shoulder abduction angles (F) from a participant with NP. RVx_n: n'th channel from a participant with NP.

The element of this vector is the angle of joints in the 3D directions at increments of 5% in the gait cycle, and the SI is constrained to be between 0 and 1. A normalized SI value of 1.0 designates an angle of zero between two vectors (i.e., the RV had an identical distribution to PRV) and signifies that motions in participants with neck pain are identical to that of the population. In **Figure 1**, the RVx is the vector with elements of individual joint angles at a specific point in the gait cycle (x%). The SI_x was computed by comparison between PRV_x and RV_x and as the cosine of angles between the two vectors. To cover the full gait cycle (0 to 100%), PRV_x and RV_x were computed at increments of 5% in the gait cycle. The SI equation was formulated between the participants with and without neck pain (where i represents 3D kinematic data/channel, and x represents the gait cycle out of 100%). Previous studies reported the mathematical description and an example of the computation of the index [27–30]. To compute a time-variant SI during gait, PRVs and RVs were computed at 5% increments of the entire gait cycle and compared between PRV and RV from corresponding segments. Therefore, the kinematic SI is a quantitative measurement of how similar a given RV is to the PRV.

As shown in **Figure 2**, the results of the study indicated that the neck pain group demonstrated a greater variation of walking patterns during the midstance and swing phases and displayed altered compensatory gait. The SI values for the gait cycle were higher in the control group than the NP group (0.98 ± 0.02 vs. 0.95 ± 0.03). The standard deviation of the SI was significantly less in the control group compared to





$$SI_x = \frac{\sum_i (PRV_{x\_i} \cdot RV_{x\_i})}{|PRV_x||RV_x|} \quad \text{Eq. 1}$$

PRVx= [PRVx_1, PRVx_1, … PRVx_n] at x% during gait
PRVx_1: averaged value of L-Shoulder flexion angles (S) from all control participants
PRVx_2: averaged value of L-Shoulder abduction angles (F) from all control participants

PRVx_n: averaged value of n'th channel from all control participants

RVx= [RVx_1, RVx_2, … RVx_n] at x% during gait
RVx_1: L-Shoulder flexion angles (S) from a participant with NP
RVx_2: L-Shoulder abduction angles (F) from a participant with NP

RVx_n: n'th channel from a participant with NP

│PRVx│= magnitude of the PRVx vector

│RVx│= magnitude of the RVx vector

Figure 1.
Example of SI computation. The left column shows the range of motions of the limbs during gait. The dark lines represent the averages of the control participants. Participant A with NP demonstrated good matching with the prototype in certain joints (R knee S, but mismatched in R hip S), and participant B with NP showed the opposite direction in L ankle abduction. On the right panel, Eq. 1 explains the principle of computing the SI. The PRVx is a vector with elements of individual joint angles at specific gate cycles, which was multiplied by averaging all control participants. (NP: neck pain, R: right, L: left, S: sagittal plane, F: frontal plane, T: transverse plane with all 24 (12 for each side) channels).

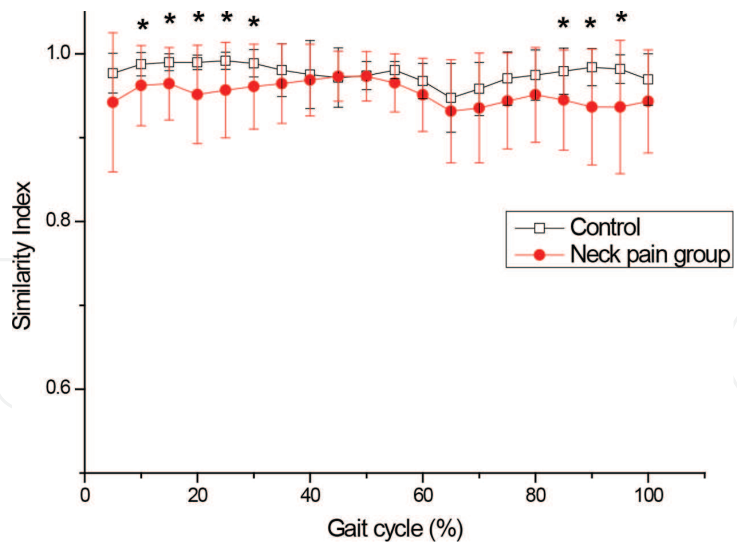


Figure 2.
The Similarity Index (SI) values were obtained from the control and neck pain groups at increments of 5% during the gait cycle. There were significant group differences at 10%, 15%, 20%, 25%, and 30% of the gait cycle in the midstance phase. In the terminal swing phase, the control group demonstrated significantly higher SI at 85%, 90%, and 95% of the gait cycle (< 0.05).*

the neck pain group (0.02 ± 0.01 vs. 0.04 ± 0.02). The similarities of the kinematic changes for the neck pain group were used to aid in the detection of limb motion differences and the resulting gait dysfunction. The SI values of the control group were significantly higher than the neck pain group during gait, especially at the midstance (10-30%) and swing (80-90%) phases. Also, the standard deviation of the SI decreased in the control group when compared to the neck pain group. The results were evident that the index reflects kinematic changes in limb functions during gait. There was less kinematic similarity in the neck pain group during gait due to a lack of similarities during the midstance and swing phases.

The neck pain group also demonstrated a greater standard deviation of the SI (ranges from 0.02 to 0.05 in the stance phase, ranges from 0.05 to 0.07 in the swing phase) compared to the control group, which was less than 0.02. The standard deviation of the SI was less in the control group compared to the neck pain group.

2.2 Clinical application of the kinematic similarity index

Although the gait parameters did not provide significant differences, the SI results detected gait deviations based on the kinematic data. The neck pain group may have modified their walking patterns, especially in the midstance and swing phases. Our results are warranted to investigate whole-body movement, including the trunk, for gait control, and early detection of gait deviations. Several studies reported that the SI provides sensitive measures as a neurophysiological method for characterizing voluntary motor control in human performance [27–31]. In our study, however, the SI concept was applied to kinematic data since the SI is a numerical expression of the similarity in distribution between participants with and without neck pain.

The importance of SI relies on the method that utilizes the kinematic data from every possible motion at a specific time point, while previous methods compared specific limb support patterns [32]. Therefore, the joint kinematic data based on the normal range (from PRV) can be detected by the SI computation. The SI concept is applicable to analyze the normalized kinematic data within and between groups for combined motions of the upper and lower limbs [27, 29], rather than for only a single joint motion in the gait cycle. Therefore, the SI is an important contribution to objective evaluations of the musculoskeletal system that could also be used to detect gait deviations.

Although our study was not intended to investigate the specific reasons for the differences, previous studies reported that the NP group could display biomechanical disturbances even with relatively mild pain and demonstrate reduced trunk rotation during gait, which indicates an increased stiffer spine [33, 34]. Previous studies were limited to clinical applications for gait dysfunction, since the distribution of the motions may provide a limited interpretation [32, 35]. A possible mechanism related to altered postural control during walking may be the consequence of diminished proprioceptive inputs leading to compensatory strategies to avoid pain or injury in the NP group [36]. The single limb support in midstance involves a progression of the body over the foot and weight-bearing stability. It is the first sub-phase where the shank rotates forward over the supporting foot, creating the second rocker motion of the cycle [37]. During the terminal swing, the final advancement of the shank takes place, and the foot is positioned for initial contact for the next gait cycle. These specific phases during gait are critical in developing effective gait strategies; however, the NP group may adapt or modify their strategies to accommodate any differences.

The differences between the SI with the combined limb motions in the midstance and swing phases during gait might be utilized to compare gait dysfunction. For example, the neck pain group may display subtle changes in load sharing and reduced conjunct neck motions in the frontal and sagittal planes during cervical rotation [38], which results in increased motion variabilities and reduced smoothness of limb motions. Therefore, the kinematic SI might be utilized to detect functional outcomes for gait dysfunction and/or balance deficits. Furthermore, clinicians need to consider gait evaluations when comparing kinematic variations for those phases, which likely reflect adaptive behavior for postural control.

Increased kinematic similarities were evident in the control group. Optimal stability and mobility might prevent potential injuries with combined trunk and limb motions [39, 40]. The results of our study also confirmed that the standard deviations of the kinematic SI decreased in the control group (**Figure 2**). The PRV was constructed from a limited set of control participants. It could be expected that individual variations and other confounding factors may provide slightly different PRVs. The SI concept would be valuable in building a database to evaluate kinematic variations between groups with and without dysfunction. There are different movement strategies that can possibly be adjusted, and the SI method detects the difference in gait dysfunctions. Further in-depth analyses could be developed for clinically meaningful and detailed insights into gait dysfunction.

3. Gait assessment between control subjects and post-stroke subjects

For post-stroke individuals, paralysis and muscle weakness in the upper and lower limbs can lead to balance, mobility disorders and gait function. There are guidelines intended for clinicians to optimize rehabilitation outcomes for subjects with chronic stroke to improve walking speed and distance. It has been generally accepted that stroke commonly results in trunk impairments that are associated with decreased trunk coordination and limited trunk muscle strength [41]. However, the guidelines limited ambulatory function due to the lack of sensitive assessments, which may not apply to the specific phase of the gait cycle in post-stroke individuals. These problems ultimately lead to a limit in functional activities and/or gait. Thus, the need for evaluation and rehabilitation methods in gait dysfunction has been emphasized on sensitive measures.

Surface EMG assessment of voluntary motor control provides a clinical manifestation of gait performance. The evidence suggests that frequency-based analysis of EMG can be used to detect cortical motor control contribution [42]. Characteristically, individuals post-stroke demonstrate slow gait velocity with residual spatial and temporal asymmetry when compared to healthy subjects. The changes in muscle activity in individuals with stroke differed between the paretic and nonparetic sides, muscle type, and gait phase; walking performance was maintained despite being affected by neuromuscular fatigue. These clinical manifestations in gait post-stroke result from deficits in limb movement as well as impairments in the control of trunk mobility [43]. Decreased anteroposterior movements of the thorax were the main variable explaining the gait function. Although trunk training is an effective strategy for improving mobility after stroke, the implementation and generalizability of this treatment approach in a clinical setting are laborious and limited. Since postural muscle activation is an integral part of motor control, it is important to investigate the activation patterns of the trunk muscles as well.

Other studies supported that neuromuscular control of gait post-stroke can be affected by changes in the trunk as well as the lower limb muscle activity patterns [41, 43]. Their impairments often result in biomechanical changes during gait. Specifically, pelvic motions might be influenced by these impairments. Following a stroke, patients walk with increased mediolateral trunk sway and larger sagittal motion of the lower trunk. Although the rotation of the upper trunk increases, the trunk shows more in-phase coordination. Acceleration of the trunk diminishes while instability and asymmetry increase as there are less movements toward the paretic side. However, it is of great importance to differentiate between compensatory trunk movements and intrinsic trunk control deficits. In **Figure 3**, the EMG analyses in the gait cycle, as averaged within the control group, were represented with solid dark lines in the first column and are regarded as the average of all strides in each muscle.

3.1 Methodology of the similarity index in gait cycle

Since the post-stroke group was not expected to have the same stance-to-swing ratio as compared with the control group even at similar gait speeds, data was normalized to eliminate this timing discrepancy. In general, post-stroke hemiparetic gait is slow compared with healthy individuals with asymmetry in the spatiotemporal parameters such as step length, swing time, stance time, and double-leg support time. Stance and swing phases were analyzed separately and then adjusted to a ratio of 60:40 percent for a graphic presentation of the gait cycle. In addition, EMG magnitude was normalized using the mean amplitude from the walking trials in a procedure similar to that reported by Benoit and colleagues [44].

The SI computation is a numerical expression of the similarity of the distribution of EMG activity in the response vectors (RV) between individuals with pathology (in this case, stroke) and a healthy population. The SI is computed as the normalized inner product, or the cosine of the solid angle between the vector representing the distribution of activity generated by healthy subjects (prototype response vector: PRV), and that representing the distribution in the test subjects (RV). Thus, the SI is constrained to lie between 0 and 1. An SI value of 1.0 designates an angle of zero (i.e., the RV had

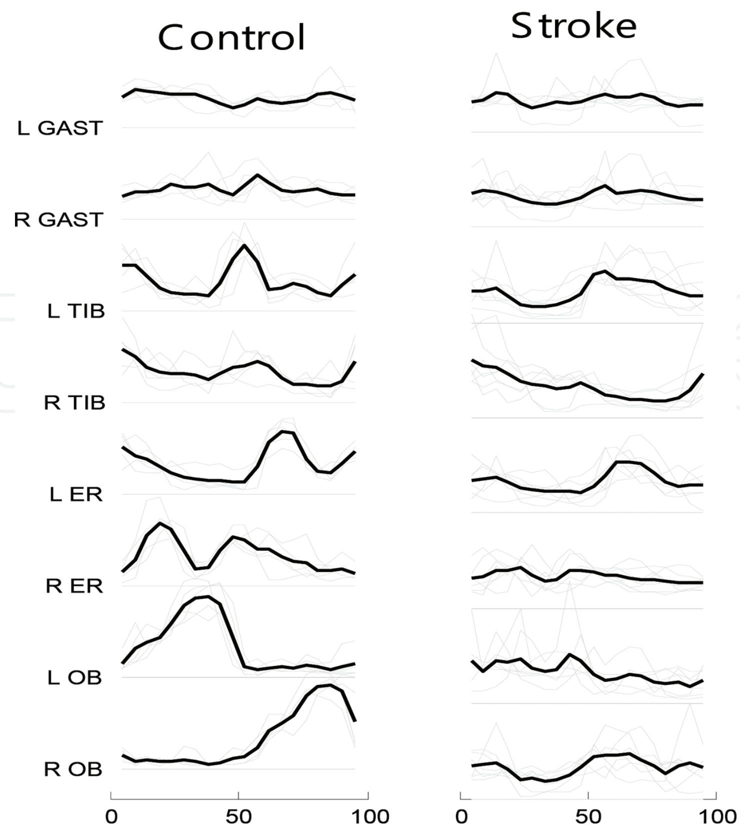


Figure 3.
EMG during gait cycle for control and stroke groups beginning at right foot contact (0%) to foot contact on the same side (100%). Individual control group EMG recordings, superimposed over the average with gray lines, demonstrated consistent patterns with small variability while individual EMG recordings for the stroke group had larger variability. This qualitative analysis of EMG patterns between the control and stroke groups suggests the spatial and temporal EMG distributions quantified by the SI. The control group demonstrated less variability than post-stroke individuals. Solid lines represent the average of all strides in each muscle from the group (EMG: electromyography, R: right, L: left, GAST: gastrocnemius, TIB: tibialis anterior, ER: erector spinae, OB: external oblique muscles).

an identical distribution of EMG activity to PRV) and signifies that muscle activity in pathological subjects is similar to that of a normal population.

The SI was computed only for RVs with a magnitude sufficient to differentiate a response from background noise. To compute a time-variant SI during the gait cycle, RVs were computed using a gait cycle segmented into 5% increments and compared to the PRVs from corresponding segments of the cycle. A gait cycle EMG, as averaged within the control group, is represented with solid dark lines in the first column of **Figure 1** and is regarded as the average of all strides in each muscle. Individual control group EMG recordings, superimposed over the average with gray lines, demonstrated consistent patterns with small variability while individual EMG recordings for the stroke group had larger variability. This qualitative analysis of EMG patterns between the control and stroke groups suggests the spatial and temporal EMG distributions quantified by the SI.

3.2 Clinical application of the similarity index during gait

This index explored the use of the SI as a method of quantifying changes in muscle activity as measured by EMG. Use of the SI indicated greater variation from the trunk muscles in normal subjects as compared to the lower limb muscles during gait in the post-stroke group. In addition, our results provided preliminary data related to trunk muscle activity in the lumbar erector spinae and the external oblique muscles, which is an area that has not been extensively reported during gait in the post-stroke group. According to the data from our lab, these differences in trunk muscle activity post-stroke peaked at 40% of the gait cycle, or 67% of the stance phase, and 90% of the gait cycle, or 75% of the swing phase. Trunk muscle activity post-stroke was most normal at 5% and 65% of the gait cycle, or 8% of the stance phase and 4% of the swing phase, which are the beginning portions of each phase.

In **Figure 4**, the SI values were obtained from the control and stroke groups without phase normalization for each 5% segment of the gait cycle. The mean of the SIs calculated for a whole gait cycle from the control group (0.965 ± 0.030) was higher than the post-stroke group (0.88 ± 0.077) when all muscles were used. The largest difference between the two groups was observed at 85-95% of the gait cycle, which would represent the middle of the swing phase if normal stance/swing phase durations were assumed. The post-stroke group showed much greater variability than the control group at all points in the cycle. The largest variability of the SI values in the post-stroke group was observed at 55% of the gait cycle. In the control group, the mean SI of each 5% segment throughout the entire gait cycle ranged from 0.965 to 0.968 with a variance of less than 0.20, which supports that the prototype represents the EMG patterns of the control group quite well.

Although the SI value from the total gait cycle for each of the two groups was significant, the SI was compared with and without phase normalization, as well as separately for the trunk and lower extremity muscles, to compare the difference between muscle groups. The SI values computed for the trunk muscles only (0.965 ± 0.034 for the control and 0.86 ± 0.10 for the stroke group) showed a difference (**Figure 4**b). However, the SI from the lower limb muscles (0.968 ± 0.026 for the control and 0.9533 ± 0.054 for the stroke group) did not show a difference between the two groups (**Figure 4**c). Therefore, the trunk muscles appear to be the major contributor to the SI difference evident between the two groups.

Due to the discrepancy of the stance-to-swing ratio between groups, the SI from the transition phase of the gait cycle between stance to swing (50-60%) had higher variability than at any other point in the cycle (**Figure 4**a). To eliminate this discrepancy, stance and swing phase timing was normalized to a ratio of 60:40 percent in the gait cycle, and the SI values were compared. When all muscles were contributing, the SI value of the control group was 0.959 ± 0.028 and 0.892 ± 0.066 for the stroke group. As expected, the SI of the control group did not change after timing normalization since phase timing is quite stable in healthy adults (**Figure 5**a as compared to **Figure 4**a). However, the SI from the post-stroke group had less variance at 50-70% of the gait cycle, and it significantly decreased at 40% and 90% of the gait cycle in comparison to the control group.

The contribution of the trunk muscles (**Figure 5**b) to the SI difference between groups was significant as compared to the lower limb muscles (**Figure 5**c). The SI of the trunk muscles during the entire gait cycle was 0.969 ± 0.029 for the control group and 0.871 ± 0.104 for the stroke group. The high variability of the SI from the stroke group was observed during the middle of the swing (85-90%) and stance (40%) phases. After phase normalization, SI values at the transition phase (50-75%) from stance to swing increased by 7% while variability decreased by 39% (**Figure 5**b). This variability, caused by a discrepancy in the swing-stance phase, would have increased by 64% if phase normalization had not been applied. The SI value from the lower extremity muscles did not demonstrate any difference between the two groups (0.970 ± 0.026 for control and 0.958 ± 0.038 for stroke). Therefore, phase normalization did

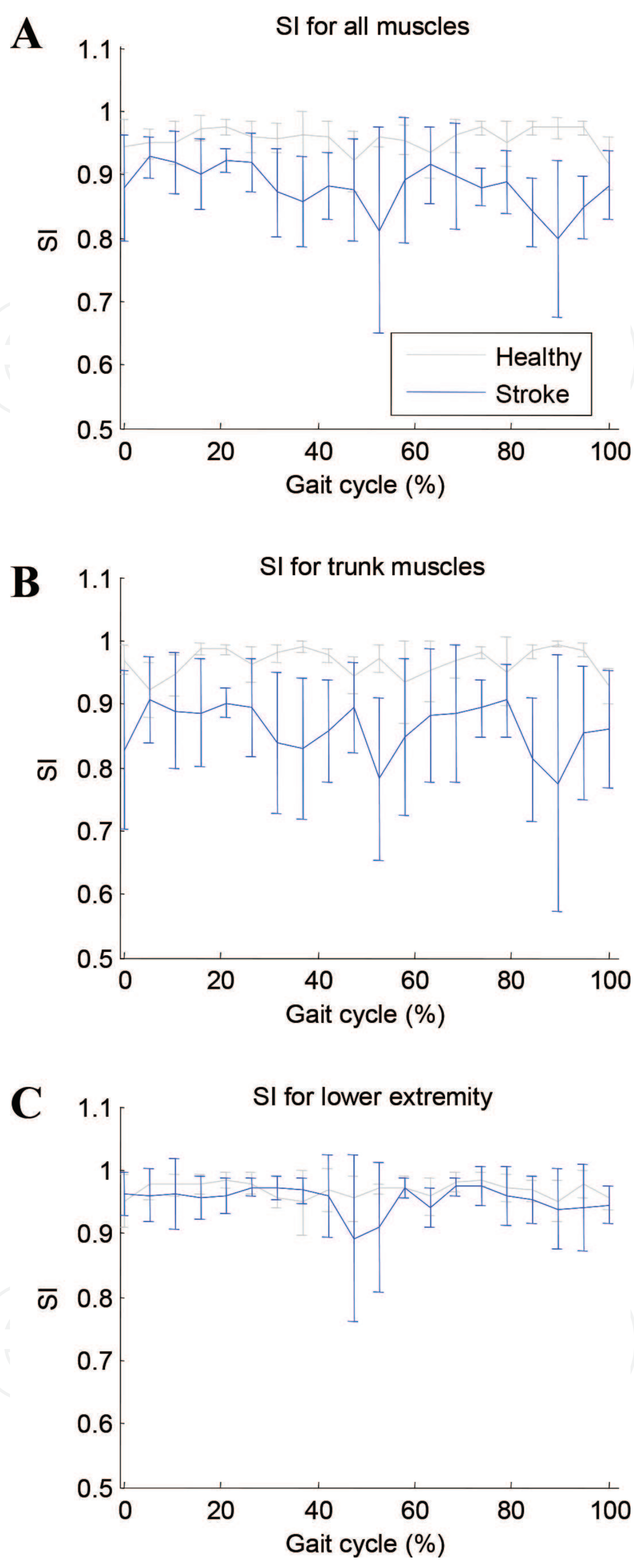


Figure 4.
The SI between control (gray line) and stroke (solid line) groups without phase normalization. The data reveal comparisons between all muscles of the stroke group and the control group (A). The SI comparing back (B) and lower limb muscles (C) indicated that the stroke group revealed increased variability. Although the individual stride EMG from the control group demonstrated consistent patterns with small variability, the stroke group demonstrated larger variability than the control group.

not change any trends seen in the non-timing, normalized SI during the gait cycle, except that the SI variability of the lower extremity muscles (**Figure 5c**) significantly decreased by 50-60% of the gait cycle.

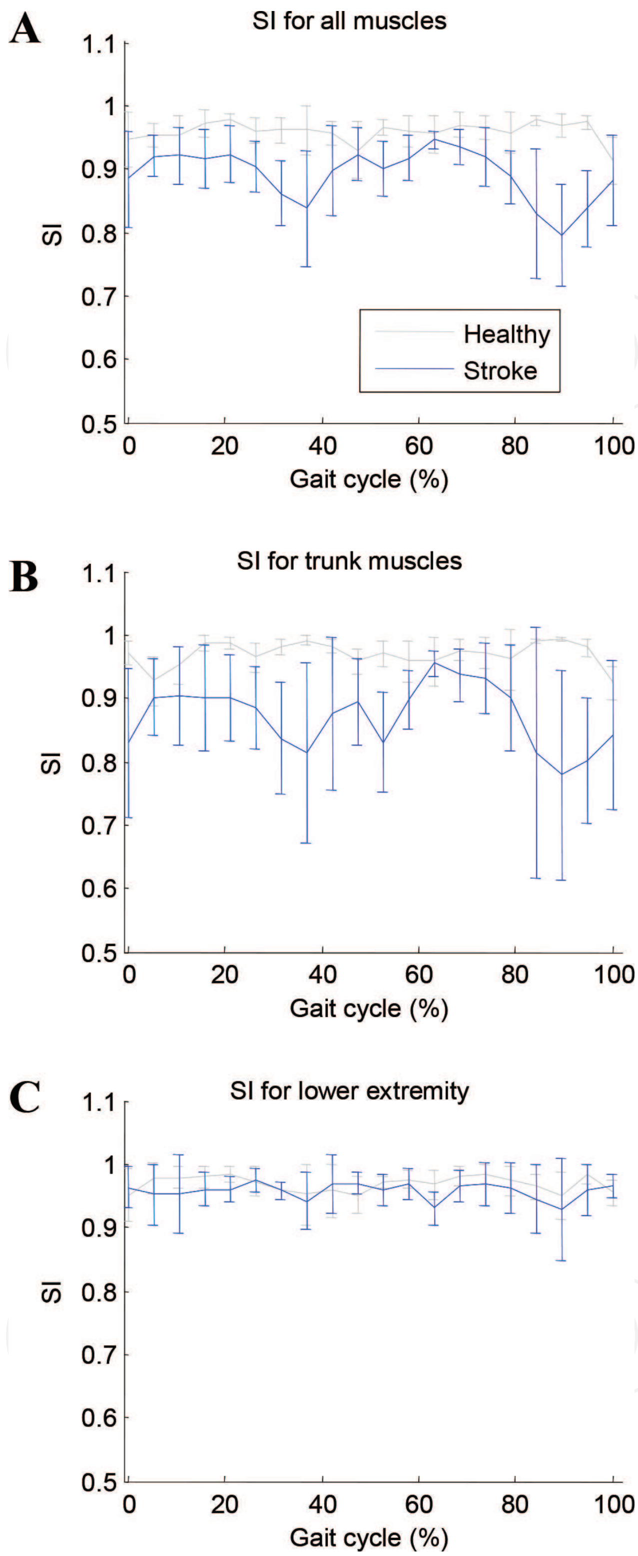


Figure 5.
The SI between control and stroke groups with phase normalization. The prototype, or control group, SI value was close to 1 with less variance in the healthy group than the post-stroke group (A). The data from trunk (B) and lower limb muscles (C) indicated that the stroke group revealed an increased SI value during the transition from stance to swing. Overall, the SI value from the stroke group had less variance at 50-70% of the gait cycle, and it significantly decreased at 40% and 90% of the gait cycle compared to the control group.

Therefore, these results indicated that phase normalization improves the capability to identify the variability between post-stroke subjects and control subjects. A loss of trunk control during walking may result from a reduction in the strength of trunk musculature, especially on the paretic side. It has been suggested that deficits in muscle strength result in reduced mobility of the pelvis in subjects with hemiparesis, which may be a protective strategy to avoid loss of balance. Individuals post-stroke also demonstrate a slower gait velocity with accompanying residual spatial and temporal trunk asymmetry when compared with the gait in healthy adults. Although the most important factors affecting gait velocity and asymmetry remain unknown, lower limb weakness is an important contributing factor. Therefore, trunk muscle activity needs to be carefully evaluated in post-stroke individuals in static and dynamic positions, in addition to trunk muscle activation patterns during gait performance.

However, one of the clinical problems in using EMG analysis with post-stroke patients is the difficulty in obtaining a valid maximum voluntary contraction, which is one accepted method for normalizing the magnitude of a muscle contraction. Nevertheless, previous research examining EMG post-stroke has assessed not only the gait cycle timing of muscle contraction through duration-referenced normalization [45], but also the relative magnitude of muscle contractions for individual muscle groups using some variation of mean or peak activity-referenced norms.

The SI is a reliable measurement of voluntary motor control for move-and-hold tasks [27]; however, the SI has not been used during gait analysis. In this study, timing normalization for phases of gait was necessary post-stroke due to timing variability (see **Figure 4** without normalization as compared to **Figure 5** with timing normalization) found in our participants. In normalizing EMG amplitude, several approaches were available for examining gait post-stroke. In gait analysis, however, peak and mean normalization are commonly accepted ways of examining the relative magnitude of muscle activity. According to the analysis by Benoit and colleagues [44], the peak EMG value is better able to identify pathological groups than average values.

The normalized EMG reflects both the timing of muscle activation and deactivation as well as the relative EMG amplitude of muscle activities. Although the SI itself does not depict the direction of variance from normal muscle activity, the amplitude of normalized EMG data shows the relative muscle activation and deactivation during the gait cycle. Originally, the SI was developed from a move-and-hold task. In addition, the SI has been used with neurologically impaired individuals as well as healthy adults [27, 29, 30]. The SI offers an important contribution to the development of evidence-based treatment evaluation in that the values produced are entirely objective. Therefore, the feasibility of using the SI to quantify the dynamic motor task of gait EMG post-stroke was explored in this study.

4. Summary of gait assessment by the SI

We examined the utilization of the SI as a sensitive tool for the conditions of musculoskeletal and neurologic dysfunctions. Individuals with neck pain have impairments of posture, balance, and coordination as well as sensorimotor deficits. The kinematic SI during gait was useful for clinical outcome measures to differentiate kinematic changes and to demonstrate quantified similarities in the gait cycle between subjects with and without neck pain. These compensatory motions are reflected by altered coordination and muscular control during the gait cycle.

The results of our EMG study in the post-stroke group indicated abnormalities in trunk muscle activity during both the swing and stance phases of gait, especially between the post-stroke and control groups. Although not completed as part of this analysis, the SI could be utilized to examine the differences, if any, in the anterior versus posterior trunk muscles bilaterally as well as hemiparetic to non-hemiparetic responses. The EMG patterns of the post-stroke and control groups during the gait cycle were analyzed by the SI, which computes the similarity of spatial and temporal distributions of muscle activity against the patterns from a control group.

A similar concept could be utilized between pathological and normal responses, as older adults with musculoskeletal dysfunction often have poor neuromuscular control, which may alter normal postural stability. Thus, there is a need to identify specific gait deviations to maintain balance and to develop effective, evidence-based strategies to improve balance control in individuals with balance deficits and to reduce their risk of falls. More importantly, the SI was a sensitive tool used to quantify the characteristics of EMG patterns when proper muscles were selected in subjects with neuromuscular dysfunction. Consequently, the compensatory limb motions resulted in increased motion variabilities and reduced smoothness of limb motions in the gait cycle. Therefore, the development and use of SI may provide an effective means to quantify muscle activity during gait. This, in turn, will assist in establishing effective treatment strategies for gait impairments as well as provide clinical insights into neuromuscular timing abnormalities, specifically for trunk musculature. The SI measure needs to be utilized to analyze gait dysfunction and rehabilitation strategies.

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