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Article

A Longitudinal Study of Physical Function Factors Related to Lower Limb Circumduction During Gait in Acute Stroke Patients with Hemiparesis

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Highlights

What are the main findings?

- Paretic femoral abduction and pelvic hike angles significantly decreased over time in acute stroke patients.
- Reduced femoral abduction correlated with paretic ankle motor recovery, while reduced pelvic hike correlated with paretic knee motor recovery.

What is the implication of the main finding?

- Femoral abduction and pelvic hike are distinct adaptive strategies, not a single "circumduction" pattern as often assumed.
- Compensatory movements in acute stroke should be re-evaluated as adaptive phenomena during recovery, not merely as abnormal movements.

Abstract

Circumduction gait in stroke patients, a compensatory movement involving pelvic hike and femoral abduction, increases energy cost. However, longitudinal studies on its mechanism during the acute phase are lacking. This study longitudinally investigated changes in the paretic femoral abduction angle during gait in acute stroke patients and identified related factors. Twenty-two acute stroke patients were assessed twice: at gait initiation and 10-14 days later. Gait kinematics during a 3m walk were measured using a depth sensor, and physical functions (SIAS) were evaluated. Changes were analyzed using paired t-tests and correlation analyses. Spatiotemporal parameters improved significantly. Kinematically, paretic femoral abduction ($p = 0.049$) and paretic pelvic hike ($p = 0.025$) significantly decreased, while maximum paretic knee flexion during swing ($p = 0.026$) increased. The decrease in femoral abduction correlated positively with the decrease in pelvic hike ($r=0.55$) and negatively with the improvement in paretic ankle motor function (SIAS) ($q=-0.49$). The decrease in pelvic hike correlated negatively with the improvement in paretic knee motor function (SIAS) ($q=-0.43$). These results suggest that in acute stroke patients, the recovery of paretic ankle and knee motor functions leads to a reduction in compensatory femoral abduction and pelvic hike, respectively. This

study provides insights for re-evaluating compensatory movements as an adaptive phenomenon during recovery, not merely as abnormal movements.

Keywords: circumduction; acute stroke; abnormal gait pattern; longitudinal change

1. Introduction

Stroke-related hemiparesis impairs gait through deficits in lower-limb sensation and motor control, reducing daily activity and health-related quality of life (QOL) [1–5]. Elevated metabolic cost is a key contributor, and so-called circumduction has been implicated as an inefficient yet common compensation [4–7]. Rather than a single pattern, circumduction is typically expressed through two surrogate kinematic behaviors during paretic swing: thigh (femoral segment) abduction and pelvic hike [8]. In this study, we define thigh abduction angle as the angle between the global vertical and the thigh segment, and pelvic hike angle as the angle between the horizontal and the line connecting both anterior superior iliac spines; positive values indicate elevation of the paretic hemipelvis [8]. Prior work shows both metrics are larger in persons with chronic stroke than in healthy adults during mid-swing [8].

Intervention studies targeting circumduction-related mechanics have yielded inconsistent results. Assistive ankle exosuits have reduced pelvic hike while increasing knee flexion in chronic stroke [10], whereas high-intensity stepping in subacute stroke improved speed and hip range of motion yet increased hip abduction during swing [11]; ankle-foot orthoses have shown no consistent effect on circumduction or pelvic hike [12]. Beyond interventions, mechanical perturbation of pre-swing knee flexion can increase paretic thigh abduction, possibly via abnormal rectus femoris–gluteus medius coupling [13]. Kinematic correlates also appear to vary with walking speed and impairment severity—faster walkers may show greater hip abduction, whereas slower walkers often exhibit reduced knee flexion/dorsiflexion and smaller abduction [14]; pelvic hike is frequently observed with limited paretic knee flexion or more severe paresis [15–17]. Collectively, cross-sectional evidence links circumduction surrogates to both segmental motor control and global spatiotemporal context, but causality and longitudinal behavior remain unresolved [12,15–18].

Clinically scalable motion analysis is essential to study these compensations during the acute phase. Markerless depth sensors estimate joint centers from depth images and allow low-cost, preparation-light 3D gait assessment suitable for wards and early rehabilitation [19–22]. Multiple studies—across healthy adults and neurologic populations—support acceptable agreement with optical motion capture for hip and knee kinematics relevant to stance and swing [23–25]. Our prior work further demonstrated concurrent validity for thigh abduction and pelvic hike in individuals with hemiparesis after brain injury, supporting the use of a depth sensor to track these compensations in clinical settings [26].

However, no longitudinal study has quantified how thigh abduction and pelvic hike change over time in the acute stage of stroke recovery, nor identified which segmental motor functions are associated with their attenuation [11,14–18]. Addressing this gap could help reframe these behaviors not merely as “abnormalities,” but as adaptive strategies that recede with recovery.

The objectives in this study were to quantify short-term longitudinal changes in paretic thigh abduction and pelvic hike during gait in acute stroke, and to identify their associations with lower-limb motor functions.

We hypothesized the followings: (i) Both thigh abduction and pelvic hike will decrease at follow-up compared with the initial assessment. (ii) Reductions in thigh abduction will be negatively associated with improvements in paretic ankle motor function, and reductions in pelvic hike will be negatively associated with improvements in paretic knee motor function.

2. Materials and Methods

2.1. Participants

This longitudinal observational study included patients with m admitted to the Departments of Neurology and Neurosurgery at *Sendai Medical Center* between May 2024 and September 2025. A total of 22 patients were enrolled.

The initial assessment was conducted once the patient was able to walk safely with minimal assistance, and the follow-up assessment was performed 10–14 days later, including gait and physical function evaluations.

The inclusion criteria were: 1) First-ever acute stroke patients aged 20–90 years; 2) Diagnosed with first-ever stroke (cerebral infarction or hemorrhage) by a physician based on Magnetic Resonance Imaging (MRI) or Computed Tomography (CT); 3) Ability to walk 3 meters with handrail use and minimal assistance; 4) Ability to understand verbal instructions. Exclusion criteria for patients with hemiparesis were: 1) presence of diabetic neuropathy, 2) motor paralysis symptoms in the non-paretic lower limb, 3) decreased sensory function in the non-paretic lower limb, 4) decreased or absent Achilles tendon reflex in the non-paretic side, 5) a history of bone or joint disease in the lower limbs, 6) presence of lesions in the cerebellum or brainstem, 7) symptoms of hemispatial neglect with a score of 2 or lower on the visuospatial cognitive test of the Stroke Impairment Assessment Set (SIAS), and 8) a Mini Mental State Examination (MMSE) score of 19 or lower[27–29].

All participants provided written informed consent prior to participation.

The study protocol was approved by the Ethics Committees of Tohoku University Graduate School of Medicine (approval no. 2024-1-032) and Sendai Medical Center (approval no. 24-12) and conducted in accordance with the Declaration of Helsinki.

2.2. Measuring Equipment

Gait data were collected using a depth sensor (Azure Kinect DK; Microsoft Corporation, Redmond, WA, USA), identical to the setup in our previous validation study (Figure 1a) [26].

The device captured skeletal joint coordinates at 30 Hz using the Azure Kinect Body Tracking SDK (v1.4.1).

The sensor was mounted on a tripod at a height of 1.0 m, positioned 4.5 m in front of the gait initiation point with a 0° frontal view.

The effective measurement range was 0.5–3.86 m [31].

This configuration has previously demonstrated excellent concurrent validity with optical motion capture for hip and knee kinematics ($r \geq 0.9$) under comparable conditions [30].

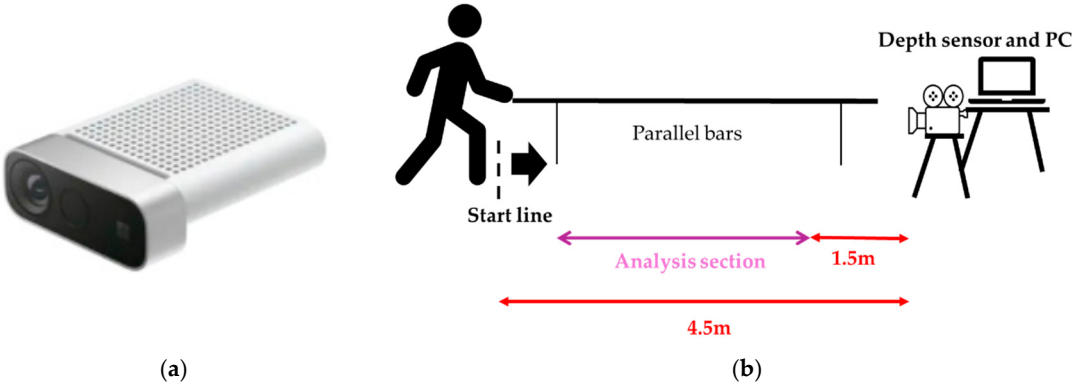


Figure 1. (a) Depth sensor; (b) Measurement environment and analysis section.

2.3. Gait Assessment

Before gait measurement, physical function was assessed using the Stroke Impairment Assessment Set (SIAS).

Participants rolled up their trouser cuffs to expose the ankle and knee joints, tucked in shirts to visualize hip movement, and walked barefoot to improve motion tracking.

From a standing start 4 m from the sensor, participants walked 3 m at a self-selected comfortable speed while lightly holding a handrail with the non-paretic upper limb (Figure 1b).

When minimal assistance was required, the evaluator used a cardboard shield to prevent self-detection by the sensor and provided minimal manual support under the paretic axilla (Figure 3).

Each participant completed five valid trials (maximum of eight, including repetitions).

Rest periods were allowed to avoid fatigue.

If a trial was invalid (e.g., tracking loss), it was immediately repeated.

To ensure consistency across sessions, handrail and orthosis use were standardized: patients who used a handrail or ankle-foot orthosis (AFO) during the initial test used the same aid during the follow-up assessment.

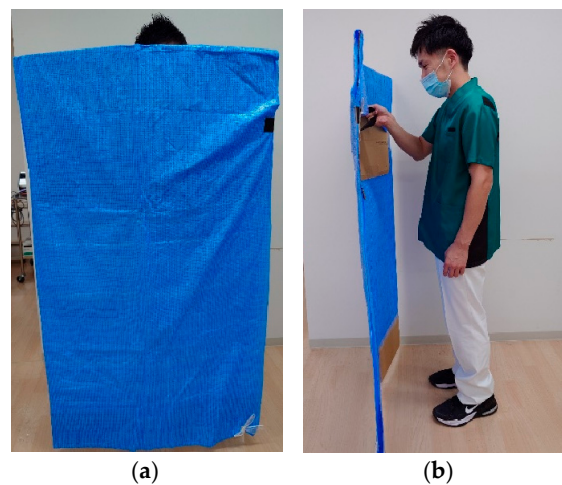


Figure 3. (a) The figure shows the evaluator holding a shielding plate as viewed from the frontal plane.; (b) This is a diagram showing the evaluator holding a shielding plate as viewed from the sagittal plane.

2.4. Data Processing

Python (version 3.10, Python Software Foundation, Wilmington, DE, USA) was used for gait analysis processing. The time-series data of each body coordinate obtained from the depth sensor were resampled to 60 Hz and smoothed using a fourth-order Butterworth low-pass filter with a 6 Hz cutoff frequency.

Gait events (initial contact and toe-off) were detected using the depth sensor data. Initial contact was defined as the point of maximum anteroposterior distance between the pelvis and ankle points. Toe-off was defined as the point of maximum anteroposterior distance between the pelvis and foot points [32].

Spatiotemporal parameters were calculated according to the methods of Rose et al. [33]. Based on previous studies [8,34], kinematic parameters were calculated, including: H1 (hip flexion angle at initial contact), H2 (hip flexion angle during loading response), H3 (minimum hip flexion angle), H4 (hip flexion angle at toe-off), H5 (maximum hip flexion angle during swing), H6 (total hip range of motion), K1 (knee flexion angle at initial contact), K2 (knee flexion angle during loading response), K3 (minimum knee flexion angle), K4 (knee flexion angle at toe-off), K5 (maximum knee flexion angle during swing), K6 (total knee range of motion), maximum paretic femoral abduction angle, and maximum paretic pelvic hike angle during 25-75% of the paretic swing phase (Figure 4).

The paretic femoral abduction angle was defined as the angle formed by the vertical axis of the laboratory and the line connecting the hip and knee joints. Paretic pelvic hike was defined as the angle formed by the horizontal axis of the laboratory and the line connecting left and right hip joints. Furthermore, a positive value was defined when the paretic hip joint was higher than the non-paretic hip joint, and a negative value when it was lower. Our previous research on the concurrent validity

between 3D motion analysis and the depth sensor in hemiplegic gait confirmed moderate or higher concurrent validity for these kinematic parameters, excluding H4, H5, and K1 [26].

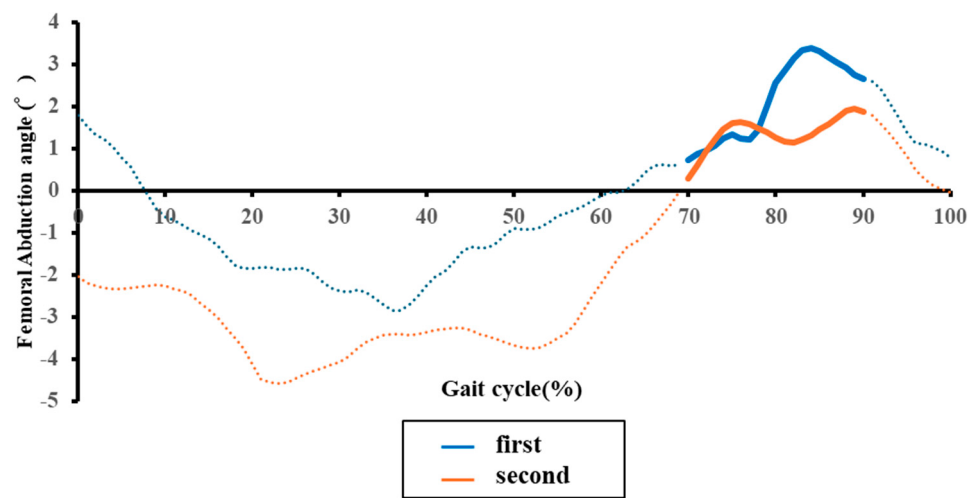


Figure 4. Representative temporal changes in the paretic femoral abduction angle during one gait cycle in an acute stroke patient. The blue and orange lines indicate data from the initial and second assessments, respectively. Solid line segments represent the mid-swing phase, while dashed line segments represent other phases.

2.5. Statical Analysis

First, the Shapiro-Wilk test was used to assess the normality of distribution for all continuous variables, including gait parameters. To examine the relationship between baseline paretic femoral abduction and pelvic hike angles and other gait and functional parameters, either Pearson's product-moment correlation coefficient or Spearman's rank correlation coefficient was used, depending on the data distribution. Comparisons of gait parameters between the initial and second assessments were conducted. A paired t-test was used for normally distributed continuous variables, while the Wilcoxon signed-rank test was used for variables that were not normally distributed. Correlations between the changes in paretic femoral abduction and pelvic hike angles and the changes in other gait and functional parameters were also analyzed using Pearson's or Spearman's correlation.

As part of the primary analysis, we calculated the effect size (Cohen's *dz*) and the correlation analysis effect size (*r*) for each comparison to quantify the magnitude of the results. These procedures were positioned as sensitivity analyses included in the primary analysis to verify the robustness and clinical significance of the results.

Correlation coefficients were interpreted as follows: 0.80 or higher as "very strong"; 0.50 to less than 0.80 as "moderate"; 0.30 to less than 0.50 as "somewhat weak"; and less than 0.30 as "weak" [35]. All statistical analyses were performed using SPSS Statistics version 20.0 (IBM, Armonk, NY, USA). The significance level was set at 0.05.

3. Results

3.1. Participant Characteristics

Twenty-two patients with first-ever acute stroke were enrolled (17 male; mean age 66.05 years, SD 15.44). The mean time from stroke onset to the initial assessment was 8.77 days (SD 5.27). The mean interval between the initial and follow-up assessments was 12.95 days (SD 1.15). Mean daily physical therapy time was 22.06 minutes (SD 7.05) before the initial assessment and 25.72 minutes (SD 7.68) during the interval period (Table 1).

Table 1. Characteristics of stroke patients.

Stroke patients		
Gender (male/female) ^a	17/5	
Age (year) ^b	66.05	(15.44)
Height (m) ^b	1.65	(0.08)
Weight (kg) ^b	68.43	(15.43)
Diagnosis (Hemorrhage/Infarction) ^a	5/17	
Paretic side (left/right) ^a	14/8	
Time since onset (days) ^b	8.77	(5.27)
Physical therapy time during longitudinal period (minutes)	25.72	(7.68)
^a Number of people		
^b Mean (Standard deviation)		

3.2. Correlations Between Compensatory Movements and Gait/Functional Parameters at Initial Assessment

At the initial assessment, the paretic femoral abduction angle showed a somewhat weak positive correlation with step width ($r = 0.46$, $p = 0.031$) (Table 2).The paretic pelvic hike angle showed moderate positive correlations with double support time ($r = 0.66$, $p = 0.001$), paretic swing time ($r = 0.57$, $p = 0.002$), and gait cycle time ($r = 0.51$, $p = 0.016$). It also showed a somewhat weak negative correlation with gait speed ($r = -0.46$, $p = 0.031$). Furthermore, paretic pelvic hike showed a moderate negative correlation with the paretic knee flexion angle at toe-off ($r = -0.50$, $p = 0.018$).Regarding physical functional assessments, the paretic femoral abduction angle showed a moderate negative correlation with paretic ankle motor function ($r = -0.56$, $p = 0.007$) and a somewhat weak negative correlation with superficial sensation ($r = -0.45$, $p = 0.034$) (Table 2). The paretic pelvic hike angle showed a somewhat weak negative correlation with hip motor function ($r = -0.44$, $p = 0.041$).

Table 2. Correlations between compensatory movements and gait parameters at initial assessment.

Spatiotemporal and kinematic variables	Paretic side femoral abduction angle		Paretic side pelvic hike angle	
	Correlation	p-value	Correlation	p-value
Gait speed (m/sec) ^a	-0.17	0.446	-0.46	0.031
Gait cycle time (sec) ^b	0.22	0.326	0.51	0.016
Double stance time (sec) ^a	0.13	0.569	0.66	0.001
Paretic stance time (sec) ^a	0.14	0.544	0.56	0.006
Paretic swing time (sec) ^b	0.42	0.053	0.57	0.002
Non-paretic swing time (sec) ^b	-0.26	0.234	-0.04	0.875
Stride length on the paretic side (m) ^a	-0.09	0.686	-0.25	0.253
Step length on the paretic side (m) ^a	0.11	0.615	-0.02	0.927
Step length on the non-paretic side (m) ^a	-0.05	0.818	-0.38	0.080
Step width (m) ^a	0.46	0.031	0.36	0.095
Paretic pelvic hike angle (°) ^a	0.33	0.131	-	-
H1 (°) ^a	0.06	0.796	-0.21	0.349
H2 (°) ^a	0.11	0.639	-0.19	0.388
H3 (°) ^a	-0.02	0.917	0.01	0.950
H4 (°) ^a	-0.09	0.698	-0.17	0.452
H5 (°) ^a	0.10	0.671	-0.19	0.391
H6 (°) ^a	0.26	0.234	-0.22	0.332
K1 (°) ^a	0.17	0.452	-0.04	0.866
K2 (°) ^a	0.34	0.119	0.11	0.636
K3 (°) ^b	-0.19	0.391	-0.10	0.669
K4 (°) ^a	-0.32	0.145	-0.50	0.018

K5 (°) ^a	-0.09	0.703	-0.35	0.108
K6 (°) ^a	-0.03	0.903	-0.28	0.210
^a Pearson's correlation coefficient (p < 0.05)				
^b Spearman's rank correlation coefficient (p < 0.05)				

Table 3. Correlations between compensatory movements and functional parameters at initial assessment.

SIAS evaluation items	Paretic side femoral abduction angle		Paretic side pelvic hike angle	
	Correlation	p-value	Correlation	p-value
Hip joint ^b	-0.10	0.654	-0.44	0.041
Knee joint ^b	-0.32	0.149	-0.38	0.078
Ankle joint ^b	-0.56	0.007	-0.40	0.064
L/E deep tendon reflex ^b	-0.20	0.372	0.02	0.939
L/E muscle tone ^b	-0.06	0.791	0.05	0.833
L/E superficial sensation ^b	-0.45	0.034	-0.11	0.634
L/E deep sensation ^b	-0.20	0.368	0.11	0.623
Ankle joint range of motion ^b	-0.05	0.822	-0.18	0.417
Abdominal strength ^b	0.17	0.444	0.01	0.957
L/E: Lower extremity				
^b Spearman's rank correlation coefficient (p < 0.05)				

3.3. Longitudinal Changes in Gait Parameters

From baseline to follow-up, spatiotemporal metrics improved: gait speed increased (p < 0.001), gait cycle time (p < 0.001), double support time (p < 0.001), and paretic stance time (p < 0.001) decreased; paretic stride length (p < 0.001) and both step lengths (paretic p = 0.003; non-paretic p < 0.001) increased (Table 4).

Compensatory angles declined: thigh abduction (mean from 6.71° to 5.79°, p = 0.049) and pelvic hike (3.82° to 2.90°, p = 0.025). Segmental kinematics improved: hip ROM (H6) (p = 0.039), maximum knee flexion in swing (K5) (p = 0.026), and knee ROM (K6) (p = 0.036) increased (Table 4).

Table 4. Changes in gait parameters over time.

Spatiotemporal parameters	Initial assessment		Second assessment		Effect size	p-value
Gait speed (m/sec) ^a	0.38	(0.16)	0.53	0.18	-1.181	< 0.001
Gait cycle time (sec) ^a	1.99	(0.68)	1.60	0.42	0.882	< 0.001
Double stance time (sec) ^b	0.49	(0.23)	0.35	0.18	0.878	< 0.001
Paretic stance time (sec) ^a	1.44	(0.54)	1.10	0.34	-0.792	< 0.001
Paretic swing time (sec) ^b	0.53	(0.16)	0.48	0.11	-0.415	0.051
Non-paretic swing time (sec) ^b	0.45	(0.08)	0.42	0.05	-0.391	0.067
Stride length on the paretic side (m) ^a	0.66	(0.15)	0.78	0.14	-1.081	< 0.001
Step length on the paretic side (m) ^a	0.31	(0.08)	0.36	0.07	-0.720	0.003
Step length on the non-paretic side (m) ^a	0.29	(0.08)	0.35	0.08	-1.022	< 0.001
Step width (m) ^a	0.13	(0.03)	0.12	0.04	0.2999	0.174
Paretic femoral abduction angle (°) ^a	6.71	(3.78)	5.79	3.47	0.513	0.049
Paretic pelvic hike angle (°) ^a	3.82	(2.34)	2.90	2.35	0.513	0.025
H1 (°) ^b	23.18	(5.67)	23.93	4.30	-0.174	0.424
H2 (°) ^a	23.20	(5.40)	24.02	4.29	-0.197	0.367
H3 (°) ^a	-3.89	(6.17)	-5.91	5.13	0.417	0.064
H4 (°) ^a	5.52	(7.12)	4.69	5.76	0.139	0.521
H5 (°) ^a	24.28	(6.79)	25.55	5.49	-0.245	0.263
H6 (°) ^a	29.59	(5.42)	32.32	5.43	-0.469	0.039
K1 (°) ^a	9.93	(5.16)	8.58	5.23	0.342	0.124

K2 (°) ^a	15.15	(4.80)	15.09	6.20	-0.010	-0.963
K3 (°) ^a	5.65	(3.16)	5.96	4.40	-0.132	0.543
K4 (°) ^a	35.24	(10.95)	37.74	8.89	-0.358	0.108
K5 (°) ^a	46.65	(11.67)	50.87	10.82	-0.513	0.026
K6 (°) ^a	41.82	(10.73)	45.84	9.77	-0.479	0.036
Mean (Standard deviation)						
^a Paired t-test (p < 0.05)						
^b Wilcoxon signed-rank test (p < 0.05)						

3.4. Longitudinal Changes in SIAS Scores

Motor scores improved at follow-up for the hip (p < 0.001), knee (p = 0.006), and ankle (p < 0.001), as did deep sensation (p = 0.046) and abdominal strength (p = 0.002). Deep tendon reflex (p = 0.153), muscle tone (p = 0.160), superficial sensation (p = 0.083), and ankle ROM (p = 0.317) showed no significant change (Table 5).

Table 5. Changes in physical function assessment over time.

SIAS evaluation items	First assessment	Second assessment	Effect size	p-value
Hip joint (0/1/2/3/4/5) ^a	1/0/6/7/8/0 (4-2)	0/0/3/3/9/7 (5-3.25)	-0.82	< 0.001
Knee joint (0/1/2/3/4/5) ^a	1/1/3/5/9/3 (4-3)	0/0/2/5/8/7 (5-3)	-0.58	0.006
Ankle joint (0/1/2/3/4/5) ^a	2/2/5/4/5/4 (4-2)	1/3/0/3/6/9 (5-3)	-0.74	< 0.001
L/E Deep tendon reflex (0/1/2/3) ^a	2/13/6/1 (2-1)	3/6/11/2 (2-1)	-0.30	0.153
L/E muscle tone (0/1/2/3) ^a	0/4/10/8 (3-2)	0/2/9/11 (3-2)	-0.30	0.160
Superficial sensation (0/1/2/3) ^a	2/0/2/18 (3-3)	0/2/1/19 (3-3)	-0.37	0.083
Deep sensation (0/1/2/3) ^a	2/0/2/18 (3-3)	0/2/0/20 (3-3)	-0.42	0.046
Ankle joint range of motion (0/1/2/3) ^a	0/0/13/9 (3-2)	0/0/10/12 (3-2)	-0.21	0.317
Abdominal strength (0/1/2/3) ^a	8/7/6/3 (2-0)	0/5/8/9 (3-2)	-0.66	0.002
SIAS: Stroke impairment assessment set (Interquartile range)				
L/E: Lower extremity				
^a Wilcoxon signed-rank test (p < 0.05)				

3.5. Correlations Between Changes in Compensatory Movements and Changes in Gait Parameters and Physical Function Assessment Items

Reductions in thigh abduction and pelvic hike were moderately coupled (r = 0.55, p = 0.008). No other gait Δ-parameters showed significant associations (Table 6).

Functionally, the reduction in thigh abduction correlated negatively with improvement in paretic ankle motor function (ρ = -0.49, p = 0.021), and the reduction in pelvic hike correlated negatively with improvement in paretic knee motor function (ρ = -0.43, p = 0.046) (Table 7).

Table 6. Correlations between changes in compensatory movements and gait parameters.

Spatiotemporal and kinematic variables	Mean (SD)	Paretic side femoral abduction angle		Paretic side pelvic hike angle	
		Correlation	p-value	Correlation	p-value
Gait speed Δ (m/sec) ^a	0.15 (0.13)	0.09	0.701	0.03	0.886
Gait cycle time Δ (sec) ^b	-0.39(0.44)	-0.05	0.809	0.23	0.299
Double stance time Δ (sec) ^a	-0.14(0.16)	-0.24	0.290	0.12	0.609
Paretic stance time Δ (sec) ^b	-0.34(0.34)	-0.19	0.393	0.10	0.661
Paretic swing time Δ (sec) ^b	-0.05(0.13)	0.32	0.148	0.41	0.058
Non-paretic swing time Δ (sec) ^b	-0.03(0.07)	-0.18	0.428	-0.05	0.828
Paretic stride length Δ (m) ^a	0.12 (0.11)	0.03	0.881	0.00	0.996
Paretic step length Δ (m) ^a	0.04 (0.06)	-0.03	0.892	0.15	0.517
Non-paretic step length Δ (m) ^a	0.05 (0.05)	0.12	0.586	-0.02	0.935

Step width Δ (m) ^a	-0.01 (0.03)	-0.22	0.332	0.12	0.604
Paretic pelvic hike Δ (°) ^a	-0.91 (1.74)	0.55	0.008	-	-
H1 Δ (°) ^b	0.75 (4.33)	-0.31	0.154	-0.32	0.148
H2 Δ (°) ^a	0.83 (4.20)	-0.34	0.118	-0.27	0.221
H3 Δ (°) ^a	-2.02 (4.84)	-0.03	0.886	-0.14	0.545
H4 Δ (°) ^a	-0.83 (5.99)	0.02	0.938	-0.14	0.540
H5 Δ (°) ^a	1.27 (5.19)	-0.13	0.557	-0.13	0.557
H6 Δ (°) ^a	2.73 (5.82)	-0.09	0.679	0.02	0.936
K1 Δ (°) ^a	-1.35 (3.94)	-0.09	0.690	-0.25	0.271
K2 Δ (°) ^a	-0.06 (5.98)	-0.23	0.311	-0.13	0.569
K3 Δ (°) ^a	0.31 (2.38)	-0.36	0.096	-0.22	0.324
K4 Δ (°) ^a	2.49 (6.96)	0.26	0.247	0.03	0.894
K5 Δ (°) ^a	4.22 (8.24)	0.10	0.660	0.09	0.700
K6 Δ (°) ^a	4.02 (8.40)	0.19	0.388	0.13	0.573

SD: Standard deviation

^aPearson's correlation coefficient (p < 0.05)

^bSpearman's rank correlation coefficient (p < 0.05)

Table 7. Correlations between changes in compensatory movements and physical function assessment items.

SIAS evaluation items	Median (IQR)	Paretic side femoral abduction angle		Paretic side pelvic hike angle	
		Correlation	p-value	Correlation	p-value
Hip joint Δ ^b	1 (1-1)	0.42	0.050	-0.20	0.371
Knee joint Δ ^b	0 (1-0)	-0.28	0.203	-0.43	0.046
Ankle joint Δ ^b	1 (1-0)	-0.49	0.021	-0.01	0.962
L/E deep tendon reflex Δ ^b	0 (1-0)	-0.34	0.121	-0.13	0.566
L/E muscle tone Δ ^b	0 (0.00)	0.01	0.969	-0.31	0.161
L/E superficial sensation Δ ^b	0 (0.00)	-0.16	0.487	-0.03	0.888
L/E deep sensation Δ ^b	0 (0.00)	0.02	0.935	0.13	0.559
Ankle joint range of motion Δ ^b	0 (0.75-0)	0.32	0.149	0.09	0.681
Abdominal strength Δ ^b	1 (1.75-0)	-0.11	0.614	-0.20	0.365

IQR: Interquartile range

L/E: Lower extremity

^bSpearman's rank correlation coefficient (p < 0.05)

Discussion

This longitudinal study quantified short-term changes in paretic thigh (femoral segment) abduction and pelvic hike during gait in acute post-stroke hemiparesis and examined their associations with segmental motor recovery. At baseline, larger thigh abduction was linked to poorer paretic ankle motor function and wider step width, whereas greater pelvic hike was linked to slower and more asymmetrical spatiotemporal patterns and to poorer knee/hip motor function. Over ~13 days, both compensatory angles decreased, and their reductions were negatively associated with improvements in ankle (for thigh abduction) and knee (for pelvic hike) motor function, respectively. Taken together, these findings support the view that these behaviors are distinct compensatory strategies that tend to recede as segmental control recovers, especially at the ankle and knee.

4.1. Relationship Between Compensatory Movements and Gait/Physical Function at Initial Assessment

At the initial assessment, greater thigh abduction co-occurred with poorer ankle motor function (SIAS) and wider step width. This pattern is consistent with a lateral stability strategy when distal control is compromised: stroke survivors have reduced accuracy of hip abduction and tend to widen step width to stabilize the stance phase [36]. In contrast, pelvic hike aligned with slower gait and

longer double-support, stance, and swing times—patterns characteristic of more severe lower-limb paresis [37]. This suggests that pelvic hike functions as a compensatory elevation to overcome functional limb shortening from limited knee flexion [38].

Little et al. classified post-stroke patients by vertical and lateral pelvic displacement and reported that those with greater displacement had poorer lower-limb motor function and greater asymmetry in hip extension and ankle plantarflexion [39]. These individuals generated less paretic propulsion in late stance, resulting in reduced knee flexion in swing and greater reliance on pelvic hike to ensure foot clearance. The present findings are consistent with this compensatory pattern observed in more severe cases, reflecting an adaptive strategy to maintain foot clearance and stance stability when paretic propulsion and knee flexion are limited.

4.2. Longitudinal Changes in Gait: Context for Compensation Attenuation

Across the cohort, gait speed, stride length, and step length increased, whereas gait cycle time, double-support time, and paretic stance time decreased. These changes coincide with improvements in hip, knee, and ankle motor function and range of motion, suggesting a transition toward more typical gait patterns as paresis recovers.

Similar trends have been reported during 2–12 months post-stroke: gait speed and step length increase alongside rising Fugl-Meyer lower-limb scores, reflecting parallel gains in walking ability and motor recovery [40,41]. Continuous gait assessments during hospitalization have also linked increased trunk acceleration regularity and improved lower-limb extension in late stance with enhanced gait speed [42,43].

However, greater speed does not always signify true paretic-side recovery—improvement may arise from non-paretic propulsion [44]. Hence, frameworks distinguishing quantitative recovery (speed gain) from qualitative recovery (paretic-side functional return) have been proposed, emphasizing paretic propulsion as a key indicator [45]. In this context, our observed increases in speed and reduction in double-support time likely mark functional progress that forms the background for the concurrent attenuation of compensatory behaviors.

4.3. Temporal Changes in Paretic Femoral Abduction Angle and Its Relationship with Physical Function

Thigh abduction decreased over time and showed a negative association with improvements in paretic ankle motor function. This can be explained by two interacting mechanisms: Firstly, propulsive recovery – Improved plantar flexor activity enhances late-stance push-off, reducing the need for lateral abduction to achieve swing-phase clearance. Secondly, coordination recovery – Abnormal reflex coupling between the rectus femoris and gluteus medius can promote excessive abduction when ankle propulsion is weak [13,47]. With better ankle control, such abnormal synergy diminishes, and lateral compensation becomes unnecessary.

Thus, the attenuation of thigh abduction reflects both mechanical sufficiency and coordination refinement—a biomechanical normalization accompanying distal recovery.

4.4. Temporal Changes in Paretic Pelvic Hike and Its Relationship with Physical Function

Pelvic hike also decreased, and its change magnitude correlated negatively with improvement in paretic knee motor control. Interestingly, the pelvic hike was more strongly related to knee control than to static flexion amplitude, implying that hike compensates not just for limited knee flexion angle but for difficulties coordinating flexion–extension timing.

Previous research showed that functional limb shortening depends on both hip and knee contributions [15], and that circumduction cannot be fully explained by reduced knee flexion or ankle dorsiflexion alone [15,17]. Akbas et al. also reported that hip circumduction is not simply a substitute for limited knee flexion, emphasizing the role of intersegmental coordination [15]. Our results—linking pelvic hike more to motor function than to angle magnitude—concur with this view.

Additionally, the depth sensor used has known proportional error in knee flexion angle [26]; this may have weakened the angle-based correlations but does not negate the stronger link to motor recovery. Altogether, improvement in knee control likely facilitates smoother foot clearance, reducing reliance on pelvic elevation.

4.5. Limitation

This study has several limitations. First, the gait measurement was conducted within parallel bars, meaning the walking distance was short and potentially influenced by handrail use. Handrail use has been reported to increase paretic lower limb propulsion [49], and gait patterns might differ under free-walking conditions. Future studies should control for the effects of gait aids or measure gait without handrails.

Second, compensatory movements of the non-paretic lower limb were not fully considered. Stroke patients may compensate for foot clearance using non-paretic ankle plantarflexion or hip extension, the extent of which could alter the magnitude of paretic-side compensation [17]. Combining quantitative evaluation of non-paretic propulsion would clarify the overall picture of compensatory movements.

Third, the physical therapy content and frequency during the observation period were not standardized. The tendency for compensatory movements to increase or decrease may differ depending on the treatment content, highlighting the need for prospective longitudinal studies that control for intervention effects.

Fourth, due to the characteristics of the depth sensor used, proportional error exists in the paretic knee flexion angles [26]. Using an optical 3D motion analysis system would improve joint angle accuracy and enable a more precise evaluation of changes over time.

Fifth, we did not perform a kinetic assessment and thus did not directly examine kinetic factors such as propulsion or ground reaction forces. While kinetic evaluation poses challenges regarding safety and equipment constraints in acute patients, future research should consider introducing simplified measurement methods using wearable sensors or force plates.

Sixth, ankle joint angles were not quantitatively assessed. It is possible that recovery of paretic ankle dorsiflexion and plantarflexion improved foot clearance, consequently contributing to the decrease in paretic femoral abduction. Future studies should conduct multi-joint analyses including ankle angles to elucidate the factors behind compensatory movements in greater detail.

5. Conclusions

This study longitudinally investigated the changes over time in paretic femoral abduction and pelvic hike angles during gait in patients with acute stroke hemiparesis and clarified the relationship between these angles and lower limb motor function.

The results showed that both the paretic femoral abduction angle and paretic pelvic hike angle significantly decreased over time, confirming a reduction in compensatory movements during gait. Furthermore, the improvement in paretic ankle motor function showed a significant negative correlation with the decrease in the femoral abduction angle, and the improvement in paretic knee motor function showed a significant negative correlation with the decrease in the pelvic hike angle.

These results suggest that in patients with acute stroke hemiparesis, the recovery of paretic ankle and knee motor functions leads to a reduction in compensatory paretic femoral abduction and pelvic hike during gait. In other words, this indicates that as the functional propulsive force and knee flexion/extension control ability of the paretic lower limb recover, the dependence on compensatory movements such as femoral abduction and pelvic hike naturally decreases.

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