Effects of Variations in Hemiparetic Gait Patterns on Improvements in Walking Speed

Scone, T.; Saadat, M.; Barton, H.; Rastegarpanah, A.

DOI:
10.1016/j.irbm.2022.08.001

IRBM ••• (••••) ••••••

ELSEVIER MASSON

Contents lists available at ScienceDirect

IRBM

journal homepage: www.elsevier.com/locate/irbm

Elsevier Masson France
EM|consulte
www.em-consulte.com

Original Article

Effects of Variations in Hemiparetic Gait Patterns on Improvements in Walking Speed

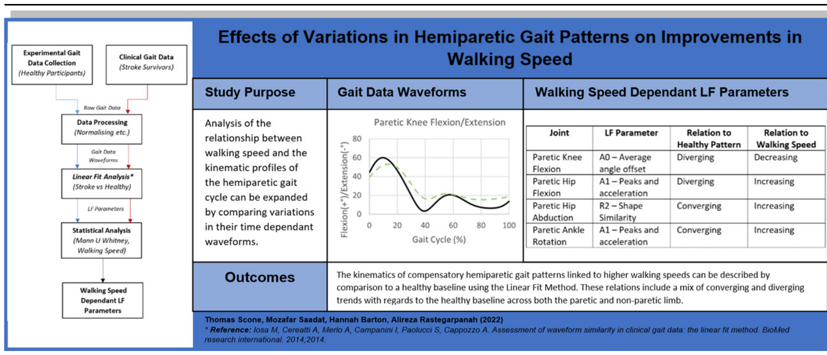
T. Scone [a,*], M. Saadat [a], H. Barton [b], A. Rastegarpanah [c]

[a] Department of Mechanical Engineering, School of Engineering, University of Birmingham, UK
[b] Department of Informatics, University Hospitals Birmingham NHS Foundation Trust, UK
[c] School of Metallurgy and Materials, University of Birmingham, UK

HIGHLIGHTS

- The linear fit method allows for in depth comparison of hemiparetic gait patterns.
- Hemiparetic gait kinematics can be described with respect to walking speed.
- Some joint kinematics diverge from healthy baselines as walking speed increases.
- Higher walking speeds can be achieved with compensatory gait techniques.

GRAPHICAL ABSTRACT



ARTICLE INFO

Article history:
Received 18 March 2022
Received in revised form 22 July 2022
Accepted 28 August 2022
Available online xxxx

Keywords:
Gait analysis
Hemiparetic gait
Stroke
Gait rehabilitation
Linear fit method

ABSTRACT

Introduction: Current research suggests that self-selected walking speed is an important indicator of hemiparetic gait rehabilitation outcome and can be targeted for improvement. Analysis of the relationship between walking speed and the kinematic profiles of the hemiparetic gait cycle can be expanded by comparing variations in their time dependant waveforms.
Methods: This paper is a pilot study to explore utilising the Linear Fit Method to compare the gait of a group of stroke survivors against a healthy baseline with respect to walking speed. This produced a set of parameters with clear physiological meaning that describe the variation of the hemiparetic gait pattern from the healthy pattern. A linear regression analysis was then performed comparing the resulting parameters against gait speed.
Results: Significant linear relationships ($p < 0.05$) were found between the Linear Fit parameters describing the hemiparetic gait pattern variations and walking speed in both paretic and non-paretic limbs. Most notably peak paretic knee flexion reduced by 20° and peak paretic hip abductions reducing to a nearly normal pattern while peak paretic hip flexions increased by 10°. The non-paretic hip flexion peak extensions remained 10° below the healthy comparison hip abduction offset was reduced but remained at nearly 2.5° to 5° from the healthy comparison.
Conclusions: As stroke survivors achieved higher walking speeds some aspects of their gait became more similar to the healthy comparison though others had no relation, or their differences became more pronounced. Combined, these relations show how paretic and non-paretic joint kinematics can be used to start identifying and quantifying effective compensatory hemiparetic gait patterns.

* Corresponding author.
E-mail address: t.scone@bham.ac.uk (T. Scone).

https://doi.org/10.1016/j.irbm.2022.08.001
1959-0318/

1. Introduction

1.1. Current compensatory gait analysis outcomes

Designing improved hemiparetic gait rehabilitation devices and programmes is an ongoing effort that often utilises gait analysis to inform methods and targets. Past studies, using a wide variety of gait analysis techniques, have focused on either identifying the limiting factors in hemiparetic gait patterns or finding aspects of the pattern that correspond with improvements of the stroke survivor's walking abilities.

Self-selected walking speed has been chosen as a parameter to assess overall gait rehabilitation effectiveness for this study. Higher self-selected walking speeds have previously been shown to correspond with overall higher ambulatory activity levels and community interaction [1–4]. This in turn has been linked to a reduction in the occurrence of post-stroke depression (PSD) [5], increases in the survivor's reported Quality of Life [6], and reducing predicted hospital costs [7] and as such is a valuable indicator of the wider outcomes of the effectiveness of gait rehabilitation.

Previously, a large focus has been placed on the asymmetry of hemiparetic gait which, being a key trait of healthy gait patterns, made it an important evaluation criterion in rehabilitation studies [8]. Balasubramanian et al. examined the relationship between self-selected walking speed and step length asymmetry and found a broad correlation between increased walking speeds and greater symmetry [9].

However, when breaking down the participants by impairment severity it was demonstrated that survivors with severe hemiparesis could still achieve high walking speeds via an asymmetric gait pattern. This leads to a more complex picture of the improvement of hemiparetic gait which becomes difficult to summarise.

Research by Kim et al. touched upon this topic with their study finding evidence of survivors achieving faster walking speeds by adopting non-normal gait patterns. These featured a prolonged hip abduction in the paretic limb which was presented as compensating for insufficient peak flexion at the paretic knee and hip joints [10]. While presenting a detailed look at walking speed and kinetic parameters such as power generation, this work used representative parameters such as the mean and range of joint angles when examining kinematic data. These representative parameters can be extremely useful when performing gait analysis but by their very nature they lead to some data, particularly time dependant data, being overlooked.

For instance, even in thorough reviews on stroke survivor's gait the kinematics of joint angles are often described in a qualitative way which is supported by specific angles pulled out at set gait pattern events [11]. This leaves it difficult to look at the exact behaviour between gait events quantitively, especially if they consist of complex waveforms, which is a barrier to more advanced forms of analysis on larger data sets, i.e. statistical.

1.2. Time inclusive gait analysis techniques

Time dependant data may instead be analysed as a waveform to preserve further data resolution. To allow for ease of comparison between trials, previous studies have taken each parameter plotted against time, e.g. spatial position or joint angle, and normalised it with regards to specific gait events [12].

Numerous methods have been suggested to compare this form of waveform data from Principal component analysis to neural networks [13,14]. These methods have the benefit of reducing information loss and can provide detailed outcomes, but these outcomes vary significantly in their ease of application and can be difficult to attribute direct physiological meaning for those not well versed in their use.

The fairly recent presentation of the Linear Fit (LF) Method as a simple and effective alternative to previous analyses brings up an interesting prospect in going back to analyse hemiparetic gait relationships in more depth [15]. The LF Method assesses the similarity between a reference waveform and the waveform of interest via three parameters that describe the shape, amplitude and offset differences in an intuitive, easy to explain manner.

The LF method considers the entirety of the gait analysis waveform data and has been shown to appropriately and reliably highlight difference in gait between healthy baseline gait and abnormal gait patterns. In addition, its ability to examine even small variations in the waveform data has been utilised in previous gait analyses and compares favourably with coefficients of multiple correlation analysis methods while maintaining easy to read physiological based outcomes [16].

The LF method has also seen use in the validation of gait analysis technologies and has proved useful in allowing for clear understanding of variations between emerging technologies and the current "gold standard" of gait analysis, the gait plug-in within the Vicon Nexus system [17,18]. Proper use of the method can thus allow an effective analysis of gait patterns down to small variations.

In this study the LF Method will be used to quantify any compensatory gait patterns linked to higher self-selected walking speeds by describing their relationship to a healthy baseline pattern.

1.3. Study aim

The aim of this pilot study is to identify any significant patterns in the 3D kinematic gait profiles of hemiparetic gait that correspond to variations in self-selected walking speed. The study will then help identify common elements of hemiparetic gait patterns that correspond to higher self-selected walking speeds. These elements could then be used to inform targets for rehabilitation therapies focusing on functional gait recovery.

2. Material and methods

2.1. Methodology overview

This pilot study took experimental gait analysis data from a group of healthy participants as a baseline gait pattern against which to compare clinical gait analysis data from a small selection of hemiparetic stroke survivors as covered in Fig. 1. The stroke survivors exhibited a range of self-selected walking speeds, a useful parameter for assessing gait functionality. This data was then prepared for further assessment by normalising each gait trial based off of key gait events.

A comparison between the two groups was made using the Linear Fit Method which compares the gait data as continuous waveforms. This method produces 3 parameters that describe how the hemiparetic gait pattern varied from the healthy reference pattern for each of the recorded gait trials.

These variations, thus quantified, were then analysed with respect to the self-selected walking speed exhibited by the participants during each gait trial. This was achieved with a Mann-Whitney U Test to establish the relationship between self-selected walking speed hemiparetic gait pattern variations.

2.2. Participants

This study used clinical gait analysis data from stroke survivors at the Gait lab of the West Midland Rehabilitation Centre (WMRC), part of Birmingham Community Healthcare NHS Foundation Trust. All stroke survivors completed a consent form informing them of

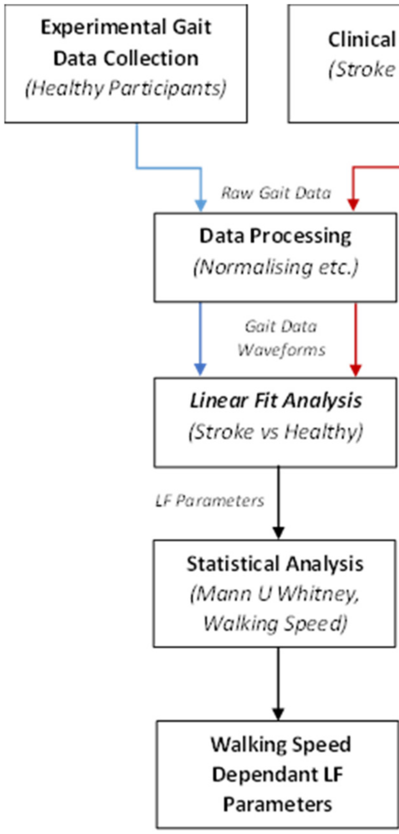


Fig. 1. Methodology flow chart showing how gait data was handled and assessed.

how their data could be used and the corresponding ethical approval was obtained by WMRC. From the data available, 4 chronic male stroke survivors were selected, all of whom had suffered right-hand side hemiparesis as the result of a stroke (age: 31 +/− 3.7 years, height 168 +/− 5.5 cm, weight: 59.8 +/− 12.4 kg). The survivors were selected as they were able to walk unaided, had achieved a range of self-selected walking speeds and had all received physical therapy after their stroke event at the WMRC.

Further gait trials were performed on a locally recruited set of healthy participants, of which 3 male participants were selected that matched the stroke survivors for gender and approximately for age (age: 25.3 +/− 1.5 years, height 172.3 +/− 5.5 cm, weight: 74 +/− 7.1 kg). All healthy participants had no history of neurological disorders or brain damage.

2.3. Experimental procedure and data processing

The system that was used consisted of a VICON MX system with 12 cameras dispersed around a central walkway for motion analysis. Of the 12 cameras, 6 were MX3+'s and 6 were MX T40's, both sets of which were capturing at 100 Hz. The VICON Nexus 1.8 Gait Analysis Software was used for capturing and processing the gait data. The cameras were calibrated at the start of each session using a Vicon "Active Wand" to ensure they were capturing correctly. The reflective markers that were used to track the participant's movements were applied according to the Vicon Lower Body marker set based off of the Newington-Helen Hayes model [19].

Participants were required to wear tight clothing in order to ensure that the marker was as close to the body as possible and stayed in position during motion. The participants were bare foot for the gait trials.

To help corroborate the data recorded by the VICON camera system and ensure processing was performed correctly, two digital cameras were used to record reference videos at 50 Hz. One was

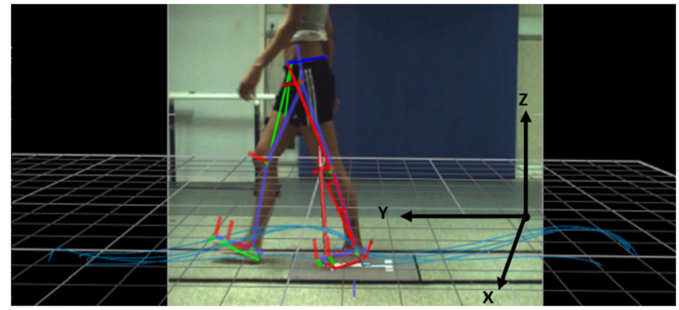


Fig. 2. View of participants walking along walkway with attached markers around the ankle, knee and hip joints with the axis labels X, Y, Z corresponding to the Frontal, Sagittal and Vertical axes of movement respectively.

placed at the end of the walkway, pointed down its length, and the other at its midpoint looking across it. Visuals from both camera sets are presented in Fig. 2.

The walkway itself was 10 m long and defined by a pair of parallel markings on the ground. Additionally, a force plate was present and outlined at the centre of the walkway as part of the standard equipment used for the gait analysis in the centre. Due to the format of the analysis, kinetic data was not included in this study.

After a series of calibration trials were completed, the participants were asked to walk along the walkway at their preferred, comfortable self-selected walking speed. The procedure followed the format specified by the WMRC for their clinical trials. The participants were made aware of the presence of the force plate but were not told to specifically target it. The presence of a clearly marked force plate has previously been found to slightly impact the step speed and length of both stroke survivors and healthy samples [20]. However, this impact is similar across both groups and no evidence of effects on kinematic parameters were found. Other studies have shown that it does not affect healthy participants gait significantly [21] and as such is deemed to present a sufficiently low risk of effecting the significance of the study outcomes.

From the processed trials, six were randomly selected from each participant for further analysis. Each of the trials was then cut down to two consecutive steps representing a snapshot of the gait pattern, right (paretic) toe-off to left (non-paretic) heel strike.

The kinematic and spatiotemporal data for the left and right hip, knee, and ankle joints were extracted as sets of continuous 3D data from which additional basic gait parameters such as step lengths and step time were also calculated. The kinematic data gave the angles at each of the joints in Flexion and Extension around the Frontal (X) axis, Abduction and Adduction around the Sagittal (Y) axis and Internal and External Rotation around the Vertical (Z) axis as highlighted in Fig. 2.

The data was then normalised to allow for effective comparisons. The paretic (right) limb toe-off event was used to mark the beginning of the gait cycle, 0%, and the non-paretic (left) limb heel-strike occurred at completion of the period, 100%. This version of the gait cycle is due to the nature of data collection employed. Collected in a clinical setting, the resulting comparable data between stroke survivors was limited while still focusing on high quality, representative data. To account for variations due to the anthropometrics the step lengths were normalised against the participant's height giving units of percentage of height per second (%h/s).

2.4. Linear fit analysis of gait patterns

The Linear Fit Method [15] was chosen to compare the gait cycle between the healthy and stroke survivors' group to allow for a straightforward comparison between the data that does not rely on comparing values at specific gait events or the minimum and maximum values.

The proposed method uses 3 formulae which compare how a continuous dataset of interest and its average (P_a, $\overline{P_a}$) the stroke survivors' gait, varies from a reference dataset and its average (P_{ref}, $\overline{P_{ref}}$), average gait of the 3 healthy participants, and produces 3 corresponding parameters. These parameters are:

a_1, or the **Amplitude Scaling Factor**, compares the rate of change between the data sets and is given by Eq. (1). It is the factor for which the reference gait data should be multiplied such that its rate of change matches the dataset of interest. I.e. if the stroke survivor data sees lower accelerations than the healthy average set then the expected value of a_1 would be <1.

$$a_1 = \frac{\sum_{i=1}^{N}(P_{ref}(\boldsymbol{i}) - \overline{P_{ref}}) \cdot (P_a(i) - \overline{P_a})}{\sum_{i=1}^{N}(P_{ref}(i) - \overline{P_{ref}})^2} \tag{1}$$

a_0, or **Scalar Addition**, is the scalar addition needed to ensure that the data of interest's value is 0 when the reference data is 0 and is given by Eq. (2). It can also be described as the average offset between the data sets.

$$a_0 = \overline{P_a} - a_1 \cdot \overline{P_{ref}} \tag{2}$$

R^2, or **Shape Similarity**, is the square of the Pearson's correlation coefficient R which establishes the strength of the linear relationship between the two waveforms that the data sets make such that their variance, or shape, matches. This is given by Eq. (3). A value of 1 is a complete match.

$$R^2 = \frac{\sum_{i=1}^{N}(a_0 + a_1 \cdot P_{ref}(i) - \overline{P_a})}{\sum_{i=1}^{N}(P_a(i) - \overline{P_a})^2} \tag{3}$$

These parameters allowed for quick and convenient comparison between the stroke survivor's gait patterns and the healthy average over the full gait cycle. If the two gait data sets were identical, the LF parameters would be $a_1 = 1$, $a_0 = 0$ and $R^2 = 1$. These parameters can thus be used to assess where hemiparetic gait differs by comparison to a healthy average gait pattern for a given gait speed.

To give a benchmark of deviations that could be accounted for within healthy gait, the healthy participants gait patterns were also analysed against the healthy average using the Linear Fit Method.

There are however limitations, as raised within the original outline of the method which highlighted that for larger variations from the shape similarity the LF method can lose definition [15] and as such care should be taken when assessing which data is significant. As suggested in the paper that originally introduced the method, if a set of joint angle waveforms had average values of R^2 below 0.50 then the data for a_0 and a_1 was not included in the discussion. Exceptions were made for cases where R^2 showed a strong positive correlation and the higher end of data was deemed comparable.

2.5. Statistical analysis of linear fit parameters

The program used to conduct the statistical analysis was IBM's SPSS Statistics [22]. The LF parameters that described the variations in the kinematic waveforms between the data set of interest and the healthy average comparison over the course of the selected gait cycle were examined using walking speed as the independent variable for the statistical analysis.

The analysis was performed in two stages, the values of R^2 of the healthy participants and the stroke survivors were directly compared to assess whether the variation from the average was significantly different and what could be accounted for by normal fluctuations in healthy gait. After testing for normality found that a significant portion of the data did not fit the parametric assumptions a Mann Whitney-U test was selected as the method to compare the groups.

The second stage was to select a test that would assess whether the variation in LF parameters was related to the Survivor's self-selected walking speed. A linear regression analysis was considered as it would effectively determine whether a linear relationship existed between the LF parameters and walking speed and its relative strength.

In order to test if the data was fit for a regression analysis a selection of scatter plots between the LF parameters and walking speed were produced using a sample taken from across various parameters and participants. Few of the plots had significant outliers and various parameters showed visual evidence of linear relationships with gait speed. An example set of scatter plots for the stroke survivors' non-paretic (left) hip are shown below in Fig. 3. The same visual analysis also looked for homoscedasticity with no clear patterns in the distribution of the data being found. Based off of the back of these preliminary analyses a Linear Regression test was found to be suitable method of analysing the LF parameters with respect to walking speed.

Statistical significance level was set at $p < 0.05$ for both the Linear Regression and Mann-Whitney U analyses.

3. Results

Additional gait parameters are presented and analysed separately to further inform the LF analysis. The average values and standard deviations of these overall parameters are given in Table 1.

The regression analysis was carried out on 3 of the gait parameters and are presented in Table 2 with their significance and regression coefficients. Gait speed was omitted as it was used as the independent variable while the cycle time was made up of the double support and step times. The healthy group showed no significant regression relationship with any of the gait parameters whereas the stroke survivors' showed relationships in the step length of both legs along with the double support time between the steps.

The LF method was applied to the kinematic waveform data for the stroke survivors with the resulting average and standard deviation for each of the kinematic LF parameters being listed in Table 3. These values describe the average differences between the stroke survivors and the healthy participants gait waveforms.

The waveforms for the kinematic joint motion are also presented in graphical form with comparisons between the healthy average waveform along with the upper and lower quartiles, Q3 and Q1 respectively, of the stroke survivors' average waveform with regards to gait speed. These graphs are contained within Figs. 4, 5 and 6 which correspond to the hip, knee and ankle joints respectively.

Within each of these 3 figures there are 6 sub-figures, each representing rotation about one axis across either the left (non-paretic) or right (paretic) joints:

- [a] = Non-Paretic Flexion/Extension
- [b] = Paretic Flexion Extension
- [c] = Non-Paretic Abduction/Adduction
- [d] = Paretic Abduction/Adduction
- [e] = Non-Paretic Internal/External Rotation
- [f] = Paretic Internal/External Rotation

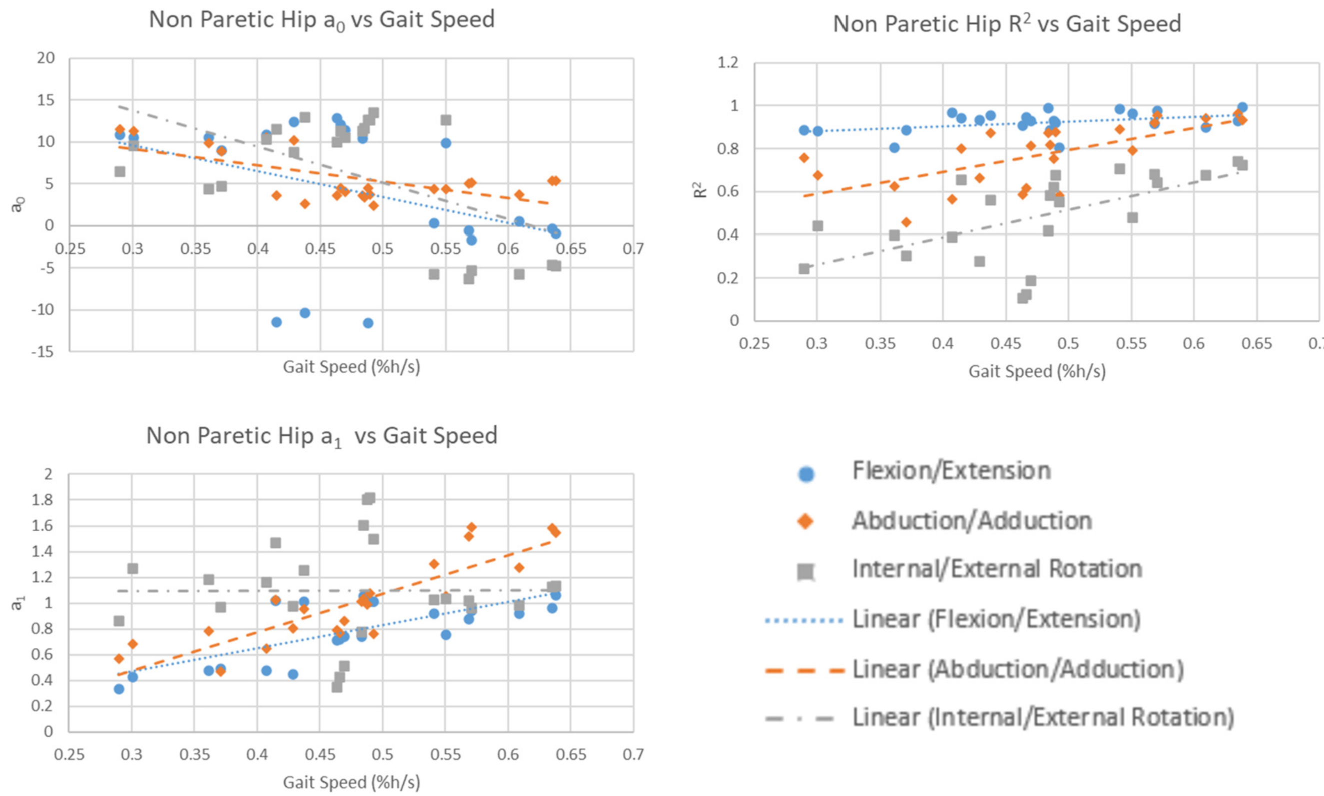


Fig. 3. Scatter plots between LF parameters and gait speed for the left (non-paretic) hip for stroke survivors.

Table 1
Average values and standard deviations of gait parameters.

Gait Parameters	Average Value (Standard Deviation)		
	Healthy	Stroke Paretic	Stroke Non-Paretic
Gait Speed (%height/s)	0.607 (.018)	0.479 (.094)	
Double Support Time (s)	0.112 (.010)	0.098 (.028)	
Step Length (%height)	32.60 (.021)	22.00 (.037)	26.90 (.040)
Step Time (s)	0.414 (.022)	0.402 (.025)	0.440 (.035)
Cycle Time (s)	0.939 (0.053)	0.940 (.063)	

Table 2
Linear regression results of basic gait parameters with gait speed as the independent variable.

Gait Parameters	p-Value (Standardised Correlation Beta Coefficient, Regression Coefficient)		
	Healthy	Stroke Paretic	Stroke Non-Paretic
Double Support Time (s)	.420 (−.203, −.107)	**.002 (−.598, −.176)**	
Step Length (%height)	.941 (−.349, −.032)	**.000 (.677, .027)**	**.002 (.594, .025)**
Step Time (s)	.156 (−.019, −.341)	.594 (.114, .030)	.198 (−.272, −.198)

To test that the two populations are significantly different, the values of R^2, waveform similarity, were compared between the stroke survivors' and healthy participants' data sets using the Mann-Whitney U test with all data split between left and right limbs, that being non-paretic and paretic for stroke survivors, and compared separately. It was found that the populations differed significantly in their waveform shape across all joints and axes in both limbs.

The regression analysis outcomes of the LF parameters of the kinematic dataset with respect to walking speed are presented in Tables 4 and 5. Table 4 contains the p-values for each LF parameter and represents where significant relationships were found. The strength of these relationships is conveyed in Table 5 with the standardised correlation coefficients and the regression coefficients. Significant regression results are presented in bold font.

The other results sections examine the LF parameters from Tables 1 and 3 and their respective correlation and regression coefficients from Tables 2, 4 and 5, focusing on those whose p-values indicate that there are significant relationships with walking speed. This is supported using the graphical waveform representations in Figs. 4, 5 and 6 where appropriate.

Table 3
Kinematic Stroke Survivor average LF parameters derived from comparison to established healthy baseline gait pattern.

Kinematic	Average LF Value (Standard Deviation)					
	Non-Paretic			*Paretic*		
Flexion/Extension (x)	Ankle	Knee	Hip	Ankle	Knee	Hip
a0	1.36° (9.98°)	2.47° (15.77°)	3.54° (6.46°)	3.84° (6.46°)	11.17° (18.84°)	4.60° (8.69°)
a1	0.784 (0.227)	0.767 (0.203)	0.752 (0.296)	0.840 (0.142)	0.709 (0.305)	0.892 (0.155)
R2	0.616 (0.209)	0.885 (0.084)	0.886 (0.183)	0.740 (0.143)	0.842 (0.158)	0.835 (0.090)
Abduction/Adduction (y)	Ankle	Knee	Hip	Ankle	Knee	Hip
a0	−2.16° (2.12°)	−1.91° (3.65°)	5.81° (2.95°)	−1.28° (1.90°)	−3.91° (6.57°)	0.89° (0.16°)
a1	0.796 (0.636)	1.136 (0.591)	0.986 (0.334)	1.197 (0.503)	1.182 (1.111)	0.954 (0.463)
R2	0.464 (0.271)	0.626 (0.272)	0.743 (0.194)	0.451 (0.280)	0.492 (0.301)	0.611 (0.230)
Internal/External Rotation (z)	Ankle	Knee	Hip	Ankle	Knee	Hip
a0	10.56° (7.31°)	−3.71° (14.26)	5.02° (9.20°)	6.51° (8.25°)	−2.86° (15.02°)	1.90° (5.36°)
a1	0.632 (0.493)	0.936 (0.366)	1.049 (0.439)	0.841 (0.420)	0.379 (0.559)	1.001 (0.464)
R2	0.450 (0.273)	0.616 (0.176)	0.466 (0.220)	0.458 (0.264)	0.228 (0.302)	0.444 (0.166)

Table 4
Linear regression analysis p-values of the LF parameters which highlight statistically significant regression between LF parameters and the independent variable, gait speed. Results below 0.05 are presented in **bold**.

Kinematic	p-Value					
	Non-Paretic			*Paretic*		
Flexion/Extension (x)	Ankle	Knee	Hip	Ankle	Knee	Hip
a_0x	.246	**.035**	.415	.471	**.019**	**.040**
a_1x	**.007**	<**.001**	**.018**	**.014**	.104	<**.001**
Abduction/Adduction (y)	Ankle	Knee	Hip	Ankle	Knee	Hip
a_0y	**.002**	**.020**	**.002**	**.011**	<**.001**	<**.001**
a_1y	**.002**	.889	<**.001**	.641	<**.001**	<**.001**
R^2y	**.006**	.274	.058	**.008**	<**.001**	<**.001**
Internal/External Rotation (z)	Ankle	Knee	Hip	Ankle	Knee	Hip
a_0z	.066	.425	**.009**	.512	.839	.167
a_1z	<**.001**	<**.001**	.747	**.004**	**.001**	<**.001**
R^2z	**.004**	.107	**.022**	**.005**	<**.001**	.098

Table 5
Linear regression analysis correlation and regression coefficients for LF parameters which describe the strength of the hemiparetic gait pattern variations from the established healthy baseline by comparison with self-selected walking speed. Results in **bold** correspond to a significant outcome ($p < 0.05$).

Kinematic	Standardised Correlation Beta Coefficient (Regression Coefficient)					
	Non-Paretic			*Paretic*		
Flexion/Extension (x)	Ankle	Knee	Hip	Ankle	Knee	Hip
a_0x	−.246 (−26.14)	**-.431 (−72.29)**	−.175 (−22.42)	−.154 (−10.60)	**-.476 (−95.36)**	**-.421 (−38.91)**
a_1x	**.532 (1.282)**	**.764 (1.648)**	**.479 (1.507)**	**.497 (.748)**	.340 (1.102)	**.772 (1.273)**
R^2x	.341 (.757)	.204 (.183)	−.020 (−.039)	.110 (.167)	.024 (.041)	**.589 (.567)**
Abduction/Adduction (y)	Ankle	Knee	Hip	Ankle	Knee	Hip
a_0y	**.596 (13.41)**	**.471 (18.27)**	**-.596 (−18.72)**	**.511 (10.32)**	**.789 (55.15)**	**.864 (23.82)**
a_1y	**.606 (4.096)**	−.030 (−.189)	**.795 (2.821)**	−.100 (−.536)	**-.828 (−9.781)**	**.689 (3.396)**
R^2y	**.542 (1.563)**	.233 (.672)	.393 (.812)	**.528 (1.573)**	**-.737 (−2.359)**	**.707 (1.729)**
Internal/External Rotation (z)	Ankle	Knee	Hip	Ankle	Knee	Hip
a_0z	−.382 (−29.69)	−.171 (−25.91)	**-.518 (−50.73)**	.141 (12.33)	−.044 (−6.98)	.292 (16.61)
a_1z	**.696 (3.650)**	**.736 (2.867)**	−.069 (−.324)	**.563 (2.516)**	**.645 (3.839)**	**-.776 (−3.826)**
R^2z	**.571 (1.659)**	−.337 (−.632)	**.466 (1.092)**	**.557 (1.561)**	**.666 (2.141)**	−.346 (−.612)

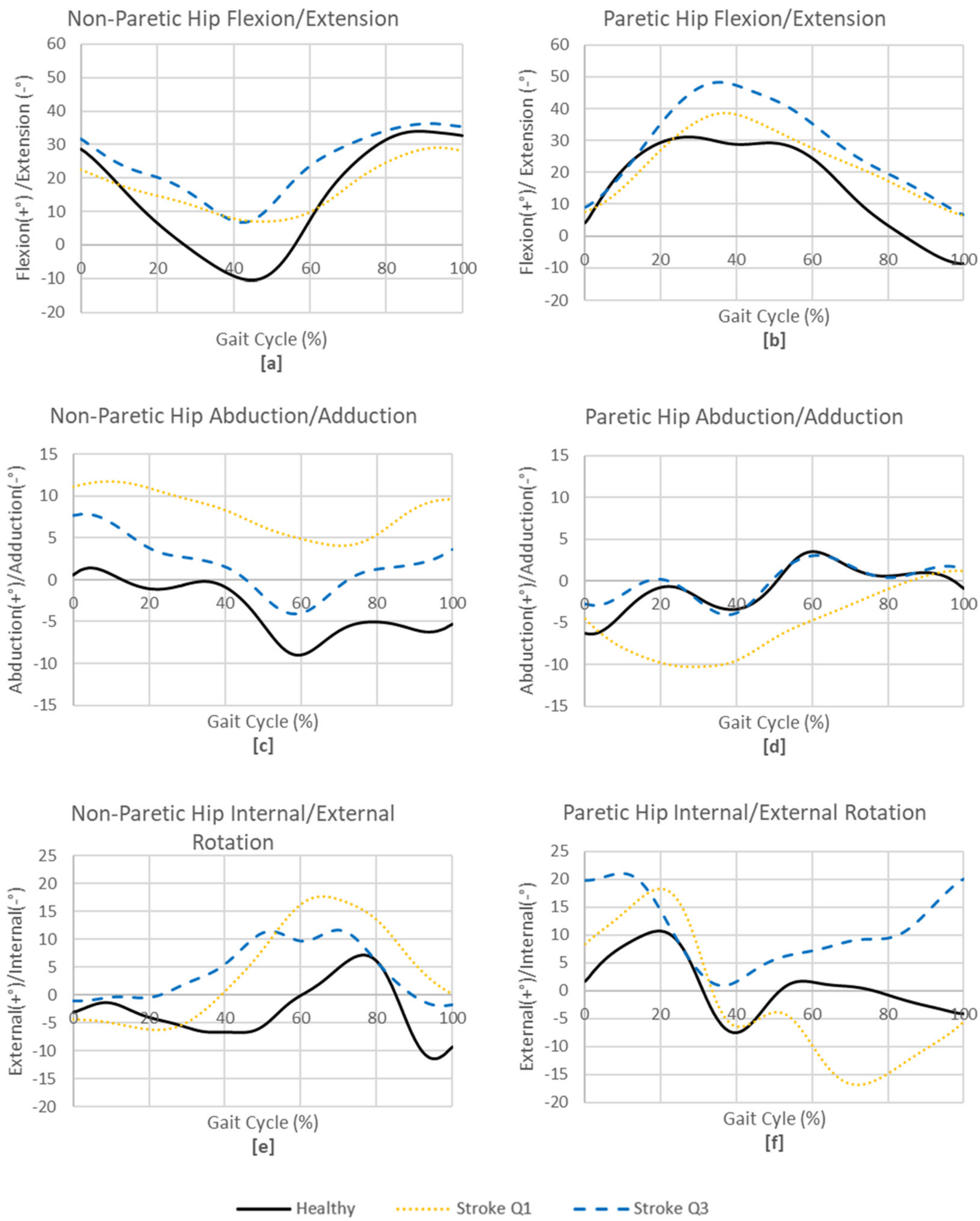


Fig. 4. Kinematic waveforms describing the variation in joint angles for the left (non-paretic) and right (paretic) hip joints over the extent of the gait trial for the average healthy gait pattern along with the lower ($Q1$) and upper ($Q3$) quartiles of average hemiparetic gait patterns for walking speed.

3.1. Step length and step time

The recorded step lengths showed a positive relationship with speed for both limbs in Table 2. The paretic limb (right) exhibited slightly higher improvements in the step length −7% more than the non-paretic limb (left) - over the range of walking speeds observed. The non-paretic leg maintained a higher stride length than the paretic leg as walking speed increased, but this gap did narrow.

The time taken per step did not see any relationship with walking speed in either the paretic or non-paretic limbs. The double support time did see an overall decrease with walking speed with a regression coefficient of −.176. The higher walking speeds were therefore accomplished due to an increase in average step length and a decrease in the amount of time between steps while the time taken for each individual step remained similar.

3.2. Circumduction in paretic and non-paretic hip joints

At slower speeds, the abduction and adduction pattern of the paretic hip had low similarity to the healthy average with a kinematic $R^2y < 0.611$. This similarity, the rate of change, a_1y, and offset, a_0y, all increased with higher speeds though, with relatively strong correlation coefficients of .707, .689 and .864 respectively. This presents a paretic hip circumduction pattern that became more similar to the healthy pattern as the self-selected walking speed increased, as can be seen in Fig. 4d.

The non-paretic hip on the other hand saw higher average similarity to the healthy comparison, $R^2y = 0.743$, for abduction and adduction but no relation to the walking speed. There was, however, a significant increase in rate of change as walking speed increased for the non-paretic hip, similar but weaker than that of

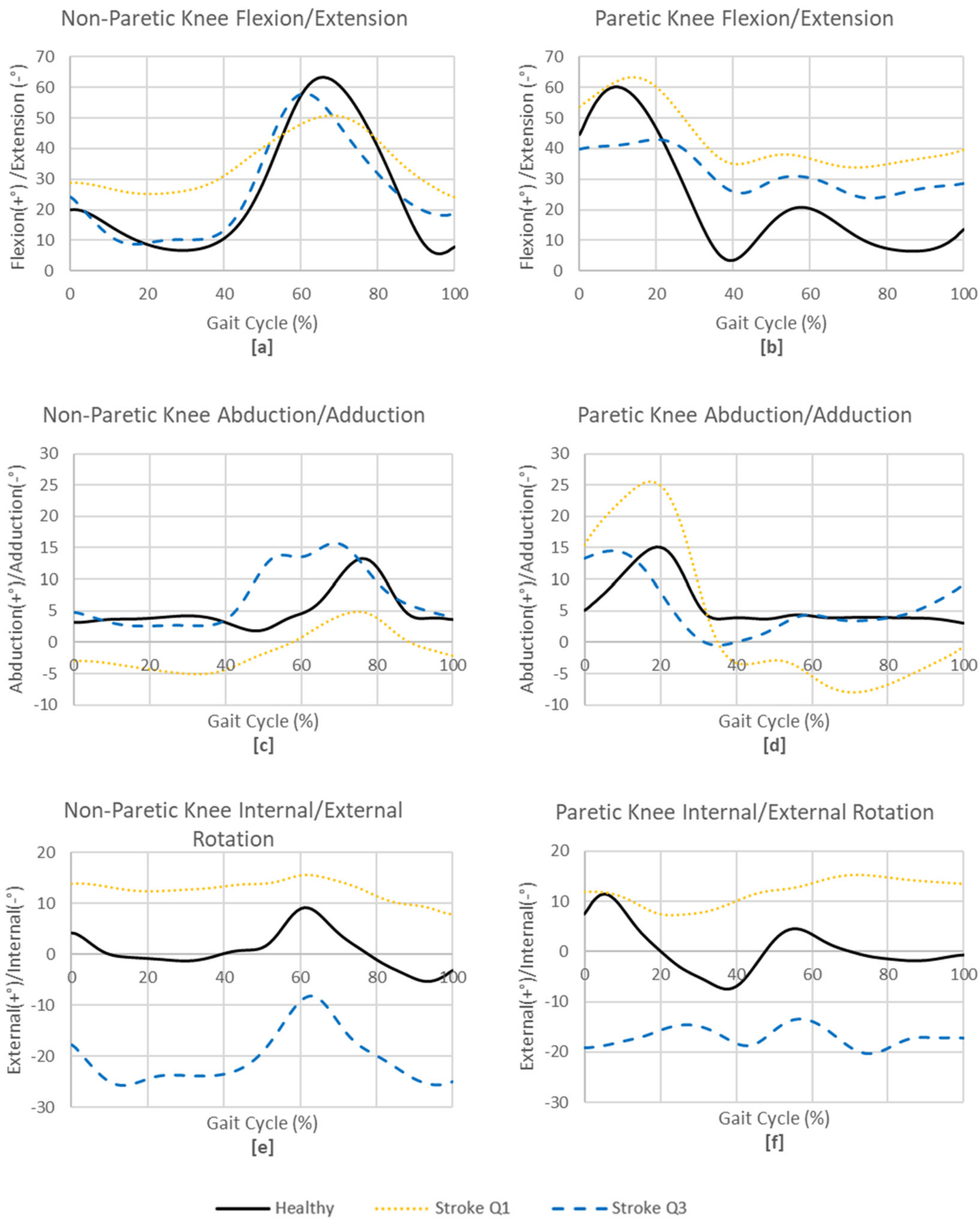


Fig. 5. Kinematic waveforms describing the variation in joint angles for the left (non-paretic) and right (paretic) knee joints over the extent of the gait trial for the average healthy gait pattern along with the lower ($Q1$) and upper ($Q3$) quartiles of average hemiparetic gait patterns for walking speed.

the paretic hip, with a regression coefficient of 2.821 for a_1y and a strong correlation coefficient of .795.

It was also noted that there was a steady decrease of average abduction in both hip joints (kinematic a_0y), increasing for the paretic hip and decreasing for the non-paretic, as gait speed increased. This led to a smaller average angle between the two limbs, which may indicate a shift in the degree of loading between limbs though the non-paretic hip was still left with a noticeable abduction offset as can be seen in Fig. 4c.

The faster walking speeds coincided with an increase in flexion of both hip joints of the stroke survivors, a_1X correlation (and regression) coefficients of .772 (1.273) and .479 (1.507) for the non-paretic and paretic hip respectively, indicating a stronger relationship for the non-paretic hip. Overall, this led to the non-paretic hip still showing reduced peak flexions as shown in Fig. 4a while the paretic hip had increased peak flexions and decreased peak extensions which can be seen in Fig. 4b.

The average waveform similarity for both hip joints was low, $R^2z \approx 0.45$, leading to difficulties in accurately assessing their relationship with walking speed. However, the value of R^2z of the non-paretic hip joint was found to correlate with increased walking speeds with a regression coefficient of 1.092 though this relationship was weaker than some others seen in the analysis with a correlation coefficient of .446. This led to a non-paretic hip joint rotation with a reduced average offset and more familiar waveform pattern when compared against the healthy baseline.

3.3. Overextension of paretic knee at higher walking speeds

One of the major causes of limb lengthening - a lower degree of flexion in the paretic knee joint - was found to be exacerbated in higher speed trials with the value of the kinematic a_0X showing a general trend towards a decrease in average flexion with regards to the healthy average (a regression coefficient of −95.36° and a correlation coefficient of −.476) as can also be seen in Fig. 5b. The acceleration of the joint angle, a_1x - and, therefore, the overall range and acceleration of motion - saw no relationship with gait speed and was lower on average than the healthy sample, mean paretic knee kinematic $a_1x = 0.709$. This data shows that the paretic knee had particularly reduced flexion during the toe-off and subsequent swing phase of the paretic leg at higher speeds (Q3), increasing its effect on limb lengthening.

The non-paretic knee also saw a decrease in overall flexion (a_0x) at faster gait speed. Unlike in the paretic leg, an increase was also observed the rate of change of flexion/extension, a_1x. This led to a relatively similar kinematic pattern of flexion and extension to the healthy comparison as can be seen in Fig. 5c.

For rotation around other axes the paretic knee had very low similarity scores, averages of $R^2y = 0.492$ and $R^2z = 0.228$, leading to difficulties in making clear comparisons. Of note though is that each similarity score did see a relationship with walking speed. While the internal and external rotation of the paretic knee joint showed a strong positive relationship with walking speed (moving closer to the healthy waveform), the Abduction/Adduction waveform actually drifted further away from the healthy pattern which can be seen to an extent in Figs. 5f and 5d respectively.

The non-paretic knee joint did not see any relation between the waveform shapes and walking speed and the similarity scores were low on average, around 0.45, with fairly large standard deviations of 0.27. This makes it difficult to compare via the statistical analysis though Figs. 5c and 5e can be used to observe the large offsets that were present.

3.4. Foot drop in paretic and non-paretic ankle joints

The analysis found that both the paretic and non-paretic ankles had surprisingly similar kinematic relationships with walking speed. Both exhibited a slight average flexion offset, $a_0x = 3.84°$ and 1.36° for the paretic and non-paretic ankle joints respectively, and large standard deviations which didn't vary significantly with walking speed.

Additionally, both ankle joints exhibited an increased rate of change with larger peak dorsiflexion angles (flexion) at higher walking speeds. For slower speeds both joints had values of a_1x below 1 indicating lower peaks than the healthy comparison but strong positive regression coefficients, particularly in the non-paretic ankle, led to the non-paretic ankle approaching near healthy flexion levels and the paretic ankle developing a distinct peak in flexion and extension as can be seen in Fig. 6a and 6b respectively. The faster paretic ankle waveform does not match the healthy waveforms shape though and instead peak extension occurs approximately 10% earlier in the gait cycle.

The rotation in other axes for both ankle joints showed lower average similarity to the healthy baseline, $R^2 \approx 0.45$, though all showed increasing similarity to the healthy pattern as walking speed increased.

The internal and external rotation in both ankle joints also showed increases in peak accelerations, a_1z, though without a shift in offset both feet remained pointing away from the sagittal plane as can be seen in Figs. 6e and 6f. The ankle joints also showed significant relationships between increasing abduction offset and walking speed, a_0y, though this remained below the healthy baseline as seen in Figs. 6d and 6c. Lastly, the non-paretic ankle joint also saw increasing peak accelerations in abduction and adduction behaviour, a_1y.

4. Discussion

This pilot study examined the evidence describing common hemiparetic gait disturbances and how they changed with respect to walking speed. The combination of these patterns shows that, while the stroke survivors adjusted their gait to compensate for the effects of hemiparetic gait as speed increased, the changes made did not end up matching the average pattern.

Instead, as the stroke survivors' walking speed increased a compensatory gait pattern emerged with peak paretic knee flexion reducing by 20° as gait speed increased between the lower and upper quartiles of gait trials while its ab/adduction pattern drifted away from the healthy baseline. It was also note there was increasing paretic ankle flexion acceleration though the pattern of flexion and extension did not relate to walking speed. The hip joint saw peak paretic abductions reducing to a nearly normal pattern while peak paretic hip flexions increased by 10°.

The non-paretic limb also exhibited compensatory elements with hip flexions never increasing from peak extensions 10° below the healthy comparison and peak ankle flexions reducing by up to 10°. The non-paretic hip also reduced the hip abduction offset but this remained at nearly 2.5° to 5° from the healthy comparison over the course of the gait cycle.

This pilot study therefore suggests that while certain aspects of the hemiparetic gait pattern may move towards the healthy baseline as walking speed increases there remains compensations in some joint kinematics that allow for these improvements.

Some detail has been lost though. The average shape similarities of the rotation in the Sagittal and Vertical axes were often fairly low and without a significant relationship being established between this value and the walking speed led to the LF parameters no longer representing the variations accurately. Potentially, a combined analysis method could prove even more effective in a future study - or a refinement of the Linear Fit Method that allowed for additional information on waveforms that deviate far away from the comparison.

It is suggested that this research may benefit future rehabilitative robotic designs which have seen a plateau in gait speed outcomes recently [22]. This may be achieved by either assisting or constraining the movement of each joint or providing training which emphasises adaption of compensatory gait patterns. Suitable robotic designs for the later method are already in development, one such design being developed in the University of Birmingham being potentially capable of both dynamic support and rendering of various surfaces [23,24].

5. Conclusion

This pilot study set out to establish the relationships between the kinematic waveforms of stroke survivors and their chosen walking speed. The Linear Fit method, along with a linear regression analysis, was able to highlight these relationships and communicate their impact on the gait pattern effectively for small to moderate variations from the healthy pattern. The results suggest that non-normal compensatory gait patterns have the potential to lead to positive rehabilitation outcomes. However, the causation and impacts of these relationships would require further study, and there is the question of the effects of long-term use of abnormal gait patterns.

Human and animal rights

The authors declare that the work described has been carried out in accordance with the Declaration of Helsinki of the World Medical Association revised in 2013 for experiments involving humans as well as in accordance with the EU Directive 2010/63/EU for animal experiments.

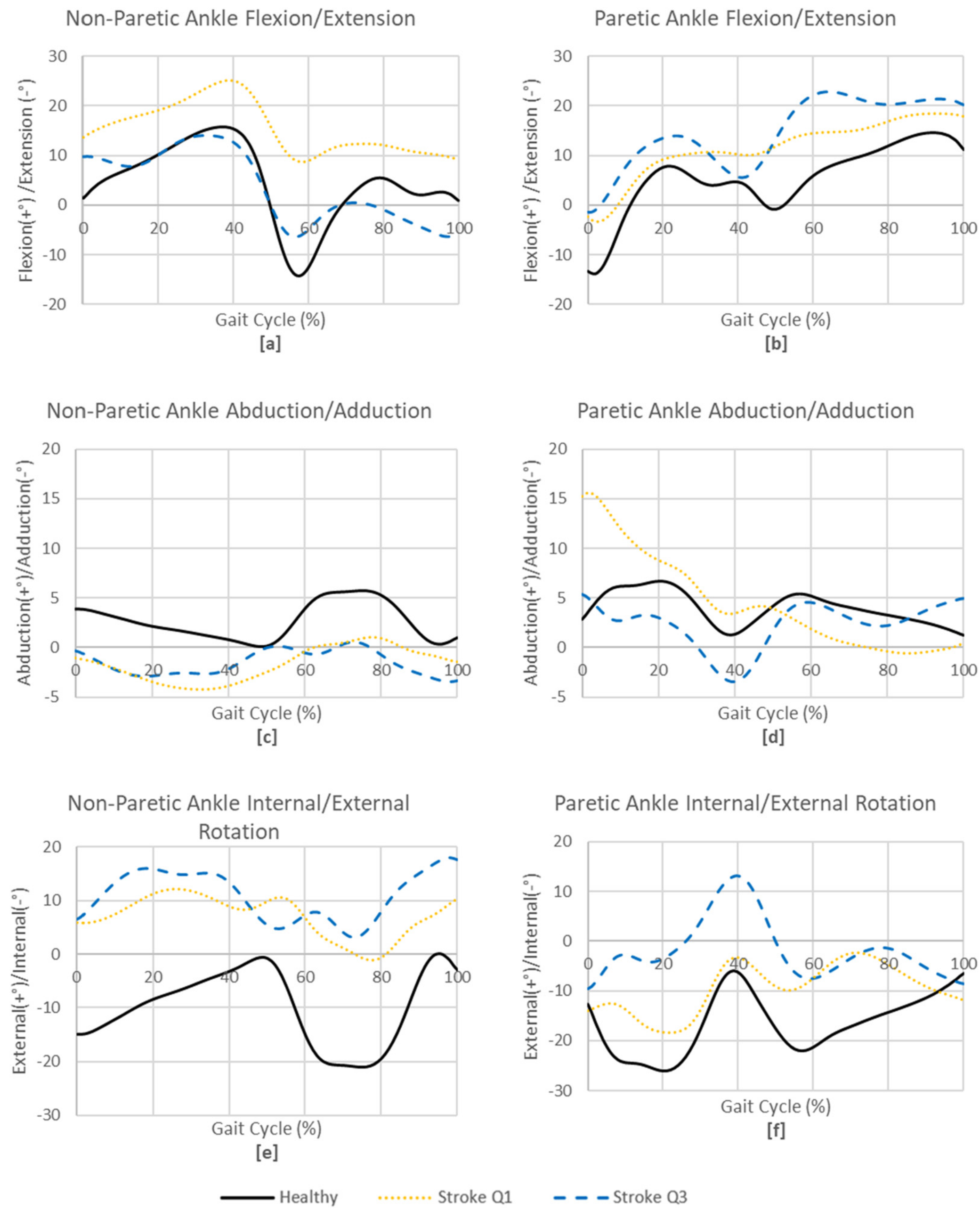


Fig. 6. Kinematic waveforms describing the variation in joint angles for the left (non-paretic) and right (paretic) ankle joints over the extent of the gait trial for the average healthy gait pattern along with the lower (Q1) and upper (Q3) quartiles of average hemiparetic gait patterns for walking speed.

Informed consent and patient details

The authors declare that this report does not contain any personal information that could lead to the identification of the patient(s).

The authors declare that they obtained a written informed consent from the patients and/or volunteers included in the article. The authors also confirm that the personal details of the patients and/or volunteers have been removed.

Funding

This work did not receive any grant from funding agencies in the public, commercial, or not-for-profit sectors.

6. Author contributions

All authors attest that they meet the current International Committee of Medical Journal Editors (ICMJE) criteria for Authorship.

CRediT authorship contribution statement

AR applied for and received the ethical approval for the data used in this study. AR performed the initial data collection and pre-processing. TS performed the literature research and conceived and performed the study's analysis. MS provided supervision during the course of the study's development. TS wrote the first draft of the manuscript with assistance and feedback of HB. All authors reviewed and edited the manuscript and approved its final version.

Declaration of competing interest

The authors declare that they have no known competing financial or personal relationships that could be viewed as influencing the work reported in this paper.

Acknowledgements

The authors thank the West Midlands Rehabilitation Centre, part of the Birmingham Community Healthcare NHS Foundation Trust for their support in the data collection used in this study.

References

[1] Fulk GD, Reynolds C, Mondal S, Deutsch JE. Predicting home and community walking activity in people with stroke. Arch Phys Med Rehabil 2010;91(10):1582–6.

[2] Bijleveld-Uitman M, van de Port I, Kwakkel G. Is gait speed or walking distance a better predictor for community walking after stroke? J Rehabil Med 2013;45(6):535–40.

[3] Michael KM, Allen JK, Macko RF. Reduced ambulatory activity after stroke: the role of balance, gait, and cardiovascular fitness. Arch Phys Med Rehabil 2005;86(8):1552–6.

[4] van de Port IG, Kwakkel G, Lindeman E. Community ambulation in patients with chronic stroke: how is it related to gait speed? J Rehabil Med 2008;40(1):23–7.

[5] Barclay R, Ripat J, Mayo N. Factors describing community ambulation after stroke: a mixed-methods study. Clin Rehabil 2015;29(5):509–21.

[6] Mayo NE, Wood-Dauphinee S, Coˆte R, Durcan L, Carlton J. Activity, participation, and quality of life 6 months poststroke. Arch Phys Med Rehabil 2002;83(8):1035–42.

[7] Husaini B, Levine R, Sharp L, Cain V, Novotny M, Hull P, et al. Depression increases stroke hospitalization cost: an analysis of 17,010 stroke patients in 2008 by race and gender. In: Stroke research and treatment, 2013; 2013.

[8] Hassid E, Rose D, Commisarow J, Guttry M, Dobkin BH. Improved gait symmetry in hemiparetic stroke patients induced during body weight-supported treadmill stepping. J Neurol Rehabil 1997;11(1):21–6.

[9] Balasubramanian CK, Bowden MG, Neptune RR, Kautz SA. Relationship between step length asymmetry and walking performance in subjects with chronic hemiparesis. Arch Phys Med Rehabil 2007;88(1):43–9.

[10] Kim CM, Eng JJ. Magnitude and pattern of 3D kinematic and kinetic gait profiles in persons with stroke: relationship to walking speed. Gait Posture 2004;20(2):140–6.

[11] Balaban B, Tok F. Gait disturbances in patients with stroke. PM&R 2014;6(7):635–42.

[12] Jacquelin Perry M. Gait analysis: normal and pathological function. New Jersey: SLACK; 2010.

[13] Chau T. A review of analytical techniques for gait data. Part 2: neural network and wavelet methods. Gait Posture 2001;13(2):102–20.

[14] Cappellini G, Ivanenko YP, Poppele RE, Lacquaniti F. Motor patterns in human walking and running. J Neurophysiol 2006;95(6):3426–37.

[15] Iosa M, Cereatti A, Merlo A, Campanini I, Paolucci S, Cappozzo A. Assessment of waveform similarity in clinical gait data: the linear fit method. BioMed Res Int 2014:2014.

[16] Horsak B, Pobatschnig B, Baca A, Greber-Platzer S, Kreissl A, Nehrer S, et al. Within-assessor reliability and minimal detectable change of gait kinematics in a young obese demographic. Gait Posture 2017;54:112–8.

[17] Al-Amri M, Nicholas K, Button K, Sparkes V, Sheeran L, Davies JL. Inertial measurement units for clinical movement analysis: reliability and concurrent validity. Sensors 2018;18(3):719.

[18] Castelli A, Paolini G, Cereatti A, Della Croce U. A 2D markerless gait analysis methodology: validation on healthy subjects. In: Computational and mathematical methods in medicine, 2015; 2015.

[19] Davis III RB, Ounpuu S, Tyburski D, Gage JR. A gait analysis data collection and reduction technique. Hum Mov Sci 1991;10(5):575–87.

[20] Rastegarpanah A, Scone T, Saadat M, Rastegarpanah M, Taylor SJ, Sadeghein N. Targeting effect on gait parameters in healthy individuals and post-stroke hemiparetic individuals. J Rehabil Assist Technol Eng 2018;5:2055668318766710.

[21] Verniba D, Vergara ME, Gage WH. Force plate targeting has no effect on spatiotemporal gait measures and their variability in young and healthy population. Gait Posture 2015;41(2):551–6.

[22] Mehrholz J, Thomas S, Kugler J, Pohl M, Elsner B. Electromechanical-assisted training for walking after stroke. Cochrane Database Syst Rev 2020;10.

[23] Maddalena M, Saadat M, Rastegarpanah A, Loureiro RC, editors. An optimized design of a parallel robot for gait training, 2017 international conference on rehabilitation robotics (ICORR). IEEE; 2017.

[24] Rastegarpanah A, Saadat M, Borboni A, Stolkin R, editors. Application of a parallel robot in lower limb rehabilitation: a brief capability study, 2016 international conference on robotics and automation for humanitarian applications (RAHA). IEEE; 2016.