



Towards Real-World Post-Stroke Gait Analysis: Iterative Development and Methodological Evaluation of a Novel, Low-Cost System

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This thesis is submitted in partial fulfilment of the requirements for the degree of
DOCTOR OF PHILOSOPHY

Declaration

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A handwritten signature in black ink, appearing to be 'R. Campbell', written over a horizontal line.

R. Campbell

Date: 20/10/2025

**"A common mistake that people
make when trying to design
something completely fool-proof is to
underestimate the ingenuity of
complete fools."**

~ Douglas Adams~

*"The Hitchhiker's Guide to the Galaxy: The Trilogy of
Five"*

Table of Contents

Declaration.....	
List of Tables	x
List of Figures	xii
List of common abbreviations	xvi
Acknowledgements.....	xviii
Abstract.....	xx
Conference Presentations Arising from this Thesis	xxii
Chapter 1 – Introduction	1
1.1 Background and Burden of Stroke.....	1
1.2 Stroke Rehabilitation and Current Assessment Practices.....	2
1.2.1 Clinical outcome measures	2
1.3 Paretic Gait and Mechanistic Insights.....	3
1.4 Measuring Gait: From Laboratory to Clinic	4
1.4.1 Three-dimensional gait analysis (3DGA).....	4
1.4.2 Shank-to-vertical angle (SVA)	4
1.5 Orthotic Practice in Stroke Rehabilitation	5
1.5.1 The role of ankle–foot orthoses (AFOs)	5
1.5.2 Muscle inhibition debate.....	5
1.5.3 Influence of design and tuning	6
1.5.4 Orthoses in integrated rehabilitation.....	6
1.6 Rationale and Thesis Focus	6
Chapter 2 – Literature Review	8
2.1 Introduction.....	8
2.2 Normal Gait and Biomechanics of Walking.....	9
2.3 Stroke and hemiplegic Gait.....	10

2.4 Measuring Hemiplegic Gait.....	13
2.4.1 Clinical outcome measures	13
2.4.2 Three-dimensional and observational gait analysis.....	15
2.4.3 Emerging hybrid approaches and service models.....	16
2.4.4 Shank-to-vertical angle (SVA) as a pragmatic marker.....	17
2.4.5 Electromyography (EMG) as a mechanistic marker	17
2.4.6 Summary of Measuring Hemiplegic Gait	20
2.5 Orthotic Practice in Stroke Rehabilitation	20
2.5.1 Theoretical concerns about muscle inhibition.....	21
2.5.2 Evidence challenging the inhibition hypothesis	22
2.5.3 Influence of AFO design and tuning	23
2.5.4 Orthoses compared with alternative interventions.....	24
2.5.5 Integration with adjunctive rehabilitation strategies	25
2.5.6 Summary of Orthotic Intervention in Stroke Rehabilitation	25
2.6 Systematic Literature Review	29
2.6.1 Introduction	29
2.6.2 Search Protocol.....	30
2.6.3 Inclusion and Exclusion Criteria	31
2.6.4 Screening and PRISMA Flow Critical Overview	33
2.6.5 Full PICO Results Table.....	37
2.6.6 Search Protocol Summary	37
2.7 Literature Review Findings	38
2.7.1 Functional Gait post-Stroke.....	39
2.7.2 Muscle Activation Patterns in Post-Stroke Gait (EMG Evidence)	44
2.7.3 Common Indicators of Gait Function	52
2.7.4 Common Practices in Stroke Rehabilitation	67
2.7.5 Rehabilitation Approaches Targeting Gait and Muscle Activity	70
2.7.6 Physiotherapy post-Stroke	72

2.8 Strength of Evidence and Risk of Bias.....	75
2.9 Conclusion and linking statement	77
Chapter 3 - Aims, hypotheses, research questions.....	80
Aim 1 - To evaluate the current practices and infrastructure for gait assessment in UK stroke rehabilitation services.	81
Aim 2 - To design and develop novel wearable technologies for the collection of gait and muscle activity data in post-stroke populations.	81
Aim 3 - To validate these devices under controlled laboratory conditions.	82
Chapter 4 – Method Overview	84
4.1 Overarching study design	84
4.1.1 Key advantages of Action research:.....	85
4.1.2 Known disadvantages with the research strategy:	86
4.2 Ontological and epistemological considerations	86
4.3 Individual Study designs	89
4.3.1 Justification of study design choice	90
4.4 Summary	95
Chapter 5 - Analysis of current assessment facilities and practices post Stroke	96
5.1 Introduction	96
5.2 Aim	97
5.3 Method.....	98
5.3.1 Study Design	98
5.3.2 Participants	99
5.3.3 Recruitment.....	99
5.3.4 Consent and Information	100
5.3.5 Survey Structure.....	100
5.3.6 Data Handling	101

5.3.7 Data Analysis	102
5.3.8 Aims-Method Alignment table	102
5.4 Results.....	103
5.4.1 Participant Characteristics.....	103
5.4.2 Stroke Patient Management.....	106
5.4.3 Use of Gait Assessment.....	107
5.4.4 Parameters Considered in Gait Assessment.....	109
5.4.5 Access to Gait Analysis.....	110
5.4.6 Orthotic Practice and AFO Use	112
5.4.7 Limitations of Gait Analysis with Orthoses	113
5.4.8 Attitudes Toward New Devices	114
5.5 Discussion	115
5.5.1 Reliance on Observational Methods.....	115
5.5.2 Limited Access to Specialist Gait Laboratories	116
5.5.3 Barriers to Detailed Assessment	116
5.5.4 Sample Size and Response Rate	117
5.5.5 Orthotic Prescription and Tuning	119
5.5.6 Appetite for Innovation	119
5.5.7 Critical Implications	120
5.5.8 Synthesis with Published Literature.....	120
5.5.9 Link to Thesis Aims	121
5.6 Summary	121
Chapter 6 – Initial design and development protocol	123
6.1 Introduction.....	123
6.1.1 Identified Clinical Needs and Gaps	123
6.2 Initial Device Concept and Design Rationale.....	126
6.3 Focussed Scoping of the Literature	127
6.3.1 Scoping Review Methodology	128

6.3.2 Scoping Review Results and Discussion	130
6.4 Design Methodology Overview	136
6.4.1 Iterative Prototyping Framework	136
6.4.2 Human-Centred Evaluation	139
6.5 Prototype Development Stages	141
6.5.1 Stage 1 — Separate Patches	141
6.5.2 Stage 2 — Integrated Patch Module	143
6.5.3 Stage 3 — Full Sock Prototype	147
6.6 Materials Selection and Specification	149
6.6.1 Fabric base with copper stitched electrode	151
6.6.2 Conductive Pathways and Electrode Zones	152
6.6.3 EMG sensor selection	152
6.6.4 IMU sensor selection	154
6.6.5 Additional Components	155
6.6.6 Safety and Contraindications	157
6.7 Summary of Prototype Development Process	158
Chapter 7 – Proof-of-Concept Cohort Case Study to validate the Novel Fabric EMG Sensor and activPAL Activity Monitor.	162
7.1 Introduction	162
7.3 Aim	164
7.4 Method.....	165
7.4.1 Study Design	165
7.4.2 Participants	166
7.4.3 Protocol	167
7.4.4 Data Analysis	175
7.4.5 Aims – Method Alignment Table	179
7.5 Results.....	181

7.5.1 Peak Shank to Vertical angle (SVA) (activPAL and Vicon comparison)	181
7.5.2 EMG	184
7.6 Discussion	189
7.6.1 activPAL versus Vicon for Peak SVA	189
7.6.2 Electromyography (EMG).....	190
7.6.3 EMG	191
7.6.4 Critical Reflection and Links to Thesis Aims.....	194
7.7 Summary	196
Chapter 8 – Feasibility study of fabric EMG sensor in novel combination with an Arduino IMU	197
8.1 Justification of Development Decisions from Chapter 7.....	198
8.1.1 Adaptation to full sock design.....	198
8.1.2 Vicon comparison.....	200
8.1.3 Justification of Final Device Design.....	213
8.2 Introduction	214
8.2.1 Ethics	215
8.3 Aim	216
8.4 Method.....	217
8.4.1 Study Design	217
8.4.2 Participants	218
8.4.3 Protocol	221
8.4.4 Data Analysis	225
8.5 Results.....	229
8.5.1 Participant Characteristics.....	229
8.5.2 Wear time and Visual Analogue Score (VAS)	232
8.5.3 SVA	236
8.5.4 EMG	246

8.5.5 Gait Speed	250
8.6 Discussion	252
8.6.1 Alignment with aims and literature context	252
8.6.2 SVA: waveform-level similarity and feature-level variability....	253
8.6.3 EMG: amplitude deficits and temporal modification.....	254
8.6.4 VAS: device comfort and usability	257
8.6.5 Gait speed: positioning the cohort against community norms and post-stroke benchmarks	259
8.6.6 Wider Orthotic Considerations Within the Stroke-Affected Cohort	261
8.7 Summary	263
Chapter 9 - Discussion.....	267
9.1 Overview of Findings	267
9.2 Integration of Findings Across Studies	268
9.2.1 Linking Clinical Need and Technological Development.....	268
9.2.2 Complementary Insights from Laboratory Validation and Real- World Feasibility.....	270
9.2.3 Feasibility Evidence from Chapter 8.....	272
9.2.4 A Coherent Narrative of Innovation and Translation.....	273
9.3 Methodological Reflection.....	276
9.3.1 Staged Research Design and Philosophical Positioning	276
9.3.2 Clinician Survey as Foundational Insight.....	277
9.3.3 Proof-of-Concept Testing under Pandemic Constraints	278
9.3.4 Feasibility Testing and the Balance of Rigour and Realism ...	280
9.4 Gait Analysis Accessibility and Equity	282
9.5 Critical Analysis of Wearable EMG/IMU Development	286
9.5.1 Novelty, Validation, and Usability	286
9.5.2 Translational Positioning and Systemic Challenges	289

9.6 Comparisons with Existing Rehabilitation Technologies.....	292
9.6.1 Observation, Laboratories, and Robotic Systems: Strengths and Systemic Failure Points Exposed by the Present Findings	292
9.6.2 Commercial IMUs and the Case for EMG–IMU Fusion: From Description to Explanation.....	295
9.7 Cost–Benefit and Health Service Implications	298
9.8 Clinical and Theoretical Implications.....	304
9.8.1 Implications for Neuro-rehabilitation Theory and Practice	304
9.8.2 Implications for Orthotic and Therapeutic Practice	306
9.8.3 Mapping Implications Back to the Central Research Question.....	308
9.9 Limitations	310
9.9.1 Technical, Methodological, and Systemic Constraints	310
9.9.2 COVID-19 Constraints.....	314
9.10 Future Possibilities.....	315
9.10.1 Technical and Analytic Refinement	315
9.10.2 Clinical Integration with Orthotic and Therapeutic Practice ..	317
9.10.3 Organisational Adoption and Workforce Development.....	318
9.11 Summary	319
Chapter 10 – Conclusion and Impact.....	322
10.1 Achievement of Research Aims.....	322
10.1.1 Summary of thesis outcomes (what was concretely delivered)	323
10.1.2 Summary of thesis conclusions (what follows from the evidence)	324
10.2 Justification of Novelty	324
10.2.1 Orthosis-compatible textile EMG integrated with inertial sensing in a single garment.....	325

10.2.2 A pragmatic, timing-dominant metric set for routine decisions	325
10.2.3 A service-centric action-research pipeline linking inequity to engineering	325
10.2.4 First steps toward an orthotic-aware evidence base	325
10.3 Impact Statement.....	326
10.3.1 Clinical practice and pathways	326
10.3.2 Patients and equity.....	326
10.3.3 Research and innovation.....	327
10.3.4 Limitations and boundaries of inference.....	327
10.3.5 Recommendations and future work.....	327
10.4 Final remarks	329
References	i
Appendix A – Individual Database Search terms	xli
A.1 PICO Table of Literature Review Papers	xliv
Appendix B – Supplementary information for Study 1.....	lxiv
B.1 University of Strathclyde Ethics	lxiv
B.2 Breakdown of Questionnaire Flow.....	lxv
B.3 Participant Information Sheet (PIS).....	lxxii
Appendix C – Supplementary information for Study 2	lxxv
C.1 Poster Presentation for ESMAC Conference, Dublin, 2022	lxxv
C.2 University of Strathclyde Ethics.....	lxxvi
C.3 Participant Information Sheet (PIS).....	lxxvii
C.4 EMG Graphs – Full Size	lxxxiii
Appendix D – Supplementary information for Study 3	lxxxviii
D.1 Poster Presentation for UK Stroke Forum, Liverpool, 2022	lxxxviii
D.2 Participant Information Sheet (PIS).....	lxxxix
D.3 Gait event detection code for Arduino system.....	xcvii

List of Tables

Table 1 - Aims / Method alignment table to demonstrate the expected outcomes mapped to individual aims for Study 1 – Questionnaire.....	102
Table 2 - Table showing the self-reported years qualified by all responders. Given in number (n=) and % delimited by less than 1 year, 1 to 4 years, 5 to 9 years and over 10 years.	104
Table 3 - Mapping of Clinical Aims, Identified Gaps, and Device Requirements	126
Table 4 - Illustrating the evolution of material and sensor selection across prototype phases. Final selections reflect alignment with device aims of low-cost accessibility, portability, and clinical relevance.....	150
Table 5 - Depicting the Aims - Method alignment and mapping to Thesis Aims from Chapter 3.	180
Table 6 - Comparison of peak mean SVA angles recorded in Vicon and with the ActivPAL sensor for all four participants walking at their comfortable, slow and fast speeds.....	183
Table 7 - Peak recorded EMG values for each participant with Wireless Delsys Trigno only, fabric EMG sensor and fabric EMG sensor plus AFO. N/A shows trials not successfully recorded due to technical issues in Vicon. * indicates recorded trials showing abnormal	185
Table 8 - Participant Demographic Data - including mean and +/- SD. Where NA = Non-Stroke Affected and SA = Stroke Affected	231
Table 9 - VAS Descriptive Statistics (Mean $\pm$ SD)	232
Table 10 - Friedman Test Results.....	233
Table 11 - Pairwise Wilcoxon Tests (Holm-corrected)	235
Table 12 - Descriptive summary of SVA features (NA, n = 8)	237
Table 13 - Normality checks (Shapiro–Wilk, α = 0.05)	237
Table 14 - Pairwise waveform similarity across NA participants	237

Table 15 - NA vs SA feature comparisons (means $\pm$ SD, Δ (SA-NA), Mann–Whitney p, Hedges' g with 95 % CIs). Provides the full numerical summary for all four features.	241
Table 16 - Feature-level comparison of AFO-restricted vs unrestrained stroke-affected participants.....	245
Table 17 - Group summaries (%RefMVC)	246
Table 18 - Cliff's delta (effect size) between groups	248
Table 19 - Reduced Participant Demographics table including Gait Speed Taken from 5-Metre Walk Test (5MWT).....	251
Table 20 - 5-Metre Walk Test - Mean and SD Gait speed by group	251

List of Figures

Figure 1 - PRISMA Flow diagram to illustrate studies included in Literature Review	35
Figure 2 - Table indicating the three individual thesis Aims mapped against the Thesis chapters in which those aims are discussed and met.....	83
Figure 3 - Circular flow diagram to illustrate the Action research strategy through Professional Practice, Critical reflection, Research, Strategic Planning and Action back around to Professional Practice (Denscombe, M. 2010). ..	85
Figure 4 - Diagram illustrating the flow of the thesis through Studies 1, 2 and 3 into the Discussion and Conclusion Chapters. Additional notes highlight the study design for each individual section.....	89
Figure 5 - Bar chart showing the self-reported number of years qualified of each responder (in years) separated into individual professions - Orthotist, Physiotherapist, Occupational Therapist, Nurse, Healthcare Assistant and Other.....	105
Figure 6 - Bar chart indicating the common settings where responders reported seeing / treating patients post-Stroke.....	106
Figure 7 - Bar chart showing the spread of responses for average time spent with patients. Under 30 minutes, 10 to 30 minutes and over 30 minutes ...	107
Figure 8 - Illustration showing Yes vs No responses for Question 6 relating to whether the respondent routinely performs a gait analysis as part of their assessment.....	108
Figure 9 - Bar chart showing the breakdown of parameters of gait used by each profession (%).....	110
Figure 10 - Pie chart to illustrate the response to question 10 - Do you feel you currently have access to suitable gait analysis for every patient?.....	111
Figure 11 - Bar chart to show responses for question 15 - What gait information do you use when prescribing or tuning an AFO?	113

Figure 12 - Illustration showing 83% positive response when asked whether a device to help ease gait analysis on patients with and without and AFO would increase regularity of gait assessment as part of routine rehabilitation practice.	114
Figure 13 - Flow diagram which shows the three stages of Fabric electrode design from separate patches to a fully integrated sock.	137
Figure 14 - Flow diagram to show the three stages involved in the device development.	138
Figure 15 - Illustration of the Human-centred evaluation loop — Showing inputs, prototype, test, feedback, iterate. (Health Foundation, 2024).....	140
Figure 16 - Image depicting the tube iteration of Prototype 2. With copper stitched EMG electrodes and button connectors visible.....	144
Figure 17 - Coronal (anterior to posterior) view of integrated tube design during testing. Delsys unit and snap leads visible anteriolaterally.	145
Figure 18 - Final prototype configuration integrating the compression sock with embedded stitched electrodes (A), MyoWare EMG board (B), Arduino Pro Mini housed in a 3D-printed anterior shank casing (C), and Phidgets FlexiForce sensor (D) positioned beneath heel.	148
Figure 19 - Image of the Prototype 1 fabric base with copper stitched EMG electrode connected to a Delsys Trigno wireless unit as used in Stage 1 testing.	152
Figure 20 - SparkFun SEN-13723 MyoWare EMG board	153
Figure 21 - Image showing Arduino Mini circuit board (IMU) connected to custom 3D printed base.	155
Figure 22 - Image of 8-inch Phidgets FlexiForce 0-100lb Resistive Force Sensor	156
Figure 23 - Image showing the full 3D printed casing for the Arduino Mini IMU unit.	158
Figure 24 - Prototype development process mapped against identified clinical aims. Shading indicates whether aims were unmet, partially met, or fully achieved at each stage.	160

Figure 25 - Diagram to illustrate placement of reflective markers on trial participant (Baudet et al., 2014)	168
Figure 26 - Image of the D-Flow software (Motek Medical, Amsterdam, Netherlands. N.d.) projecting a virtual pathway onto a front-facing screen.	169
Figure 27 - Electrode placement for the lateral gastrocnemius according to SENIAM guidelines. SENIAM. (n.d.)	172
Figure 28 - An image of a stock, 3/4 length footplate Ankle Foot Orthosis (AFO) from Chaneco, (2021).	173
Figure 29 - Bar chart showing peak gastrocnemius activation timing across participants and conditions (Trigno, Fabric, Fabric + AFO).....	186
Figure 30 - Bar chart showing the differences in peak EMG timing between the fabric sensors (with and without AFO) and the Trigno system for each participant.	187
Figure 31 - All available EMG (lateral head Left gastrocnemius muscle) for participants 1 to 4 at each data collection - Delsys Trigno EMG, Fabric EMG and Fabric EMG plus AFO.	188
Figure 32 - Illustrating pathway from individual electrode patches to a fully integrated sock design	198
Figure 33 - Image depicting full set up with FSR, MyoWare board, EMG sock, Arduinio IMU and backing plate	199
Figure 34 - Participant A: Z-score vs RIGHT SD.....	205
Figure 35 - Participant A: Bland–Altman plot for Right side	206
Figure 36 - Participant B: Bland–Altman plot	207
Figure 37 - Agreement for right-leg SVA (0–100% GC): Arduino anterior-shank IMU vs Vicon Nexus—r, RMSE/MAE, nRMSE, bias, DTW, %±1/±2 SD, peak diff & timing.	210
Figure 38 - Visual Analogue Scores (VAS) across time points for neurologically intact (NA) and stroke-affected (SA) participants: A) individual trajectories with group means, B) group means ± SD, and C) distributions with individual datapoints.	234

Figure 40 - Stroke-affected AFO-restricted vs unrestrained: difference curve with cluster-permutation inference.	243
Figure 42 - Box plot: Peak timing (%GC) by group	249
Figure 43 - Stroke vs Stroke+AFO: significant %GC ranges (FDR, %RefMVC)	250

List of common abbreviations

AFO – Ankle–Foot Orthosis

CINAHL – Cumulative Index to Nursing and Allied Health Literature

CMAS – Clinical Movement Analysis Society

CV – Coefficient of Variation

DTW – Dynamic Time Warping

EMG – Electromyography

EMBASE – Excerpta Medica Database

ESMAC – European Society for Movement Analysis in Adults and Children

FAC – Functional Ambulation Categories

FAQ – Frequently Asked Questions

FDR – False Discovery Rate

FES – Functional Electrical Stimulation

FSR – Force Sensitive Resistor

GBD – Global Burden of Disease

GC – Gait Cycle

GRF – Ground Reaction Force

IMU – Inertial Measurement Unit

MAE – Mean Absolute Error

MG – Medial Gastrocnemius

NA – Neurologically Intact

NICE – National Institute for Health and Care Excellence

NIH – National Institutes of Health

NIHR – National Institute for Health and Care Research

OGA – Observational Gait Analysis

OTS – Off the Shelf

PCA – Principal Component Analysis

PEDOT:PSS - poly(3,4-ethylenedioxythiophene):polystyrene sulfonate

PIS – Participant Information Sheet

PLS – Posterior Leaf Spring (orthosis)

PP – Paretic Propulsion

PRISMA – Preferred Reporting Items for Systematic Reviews and Meta-Analyses

RAGT – Robotic-Assisted Gait Training

RMSE – Root Mean Square Error

SA – Stroke-Affected

SD – Standard Deviation

SENIAM – Surface Electromyography for the Non-Invasive Assessment of muscles

SKG – Stiff-Knee Gait

SSNAP – Sentinel Stroke National Audit Programme

TA - Tibialis Anterior

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Abstract

Stroke is a leading cause of long-term disability, with gait impairment among its most persistent consequences. In the UK, rehabilitation services rely heavily on observational assessment, while access to specialist laboratories is limited. This creates a gap between biomechanical insight and the tools available to clinicians. This thesis addresses that gap through the development and evaluation of wearable technologies for real-world gait analysis.

The research pursued three aims: 1) to evaluate current UK gait assessment practices; 2) to design and iteratively develop orthosis-compatible measurement tools integrating electromyography (EMG) and inertial measurement units (IMUs) and 3) to validate these devices in both laboratory and clinical contexts.

A nationwide survey revealed dependence on visual methods, limited orthotic tuning, and strong demand for low-cost, portable tools. In response, a compression-sock-based system embedding textile EMG, IMU, and force sensors was created. Proof-of-concept testing showed that it could measure shank-to-vertical angle, stance-time asymmetry, and tibialis anterior activation timing with accuracy comparable to gold-standard systems. A feasibility study confirmed that clinically interpretable activation–timing data could also be obtained outside the laboratory, distinguishing recovery from compensation.

The thesis makes four contributions: the first orthosis-compatible garment to integrate textile EMG with inertial sensing; a pragmatic, timing-dominant metric set aligned with clinical decisions; a service-centred research pipeline linking inequity to engineering innovation. Together, these contributions advance the integration of mechanistic gait analysis into routine practice.

The findings demonstrate that wearable EMG–IMU systems can embed mechanistic insight into everyday orthotic prescription, enabling more precise orthotic tuning and informing debates on recovery versus compensation. By bridging biomechanics and clinical care, this work supports more equitable, data-driven pathways in stroke rehabilitation.

Conference Presentations Arising from this Thesis

2022 - ESMAC Conference, Dublin – Poster presentation (Appendix C.1)

“Preliminary validation study of novel fabric EMG device in non-affected adults – can it compare to the industry gold-standard, Vicon?”

2022 - UK Stroke Forum, Liverpool – Poster presentation (Appendix D.1)

“Preliminary validation study of novel AFO-compatible fabric EMG device in non-affected volunteers.”

2023 - ISPO World Congress, Guadalajara – Free paper presentation

“Development of tools for post-Stroke data collection - Validation of novel fabric EMG sensor, with Arduino-driven data collection, on non-affected participants”

Chapter 1 – Introduction

1.1 Background and Burden of Stroke

“Although the world is full of suffering, it is also full of the overcoming of it.”

Helen Keller, 1880-1968

Stroke remains a leading cause of death and disability worldwide, with profound individual, societal, and economic impacts. According to the Global Burden of Disease (GBD) Study, stroke is the second-leading cause of death globally and the third-leading cause of disability-adjusted life years (DALYs) lost, accounting for approximately 12 million new cases annually (GBD 2021 Stroke Collaborators, 2022). The lifetime risk of stroke has risen over recent decades due to population ageing and the persistence of vascular risk factors, with current estimates suggesting that one in four adults over the age of 25 will experience a stroke during their lifetime (Feigin et al., 2018).

Survival rates after acute stroke have improved in many countries due to advances in acute care and early management; however, these successes have shifted the challenge from mortality to long-term disability. Up to 80% of stroke survivors experience motor impairments, and approximately 50% continue to experience persistent walking limitations (Lawrence et al., 2001; Winstein et al., 2016). The restoration of safe and efficient ambulation is therefore among the highest priorities for patients, clinicians, and healthcare systems alike.

The global economic burden is substantial. In Europe, the combined direct and indirect costs of stroke were estimated at €60 billion in 2017, with informal care and productivity losses constituting almost half of the total (Luengo-Fernandez et al., 2020). In the UK, stroke accounts for approximately 5% of total NHS

expenditure, with costs projected to increase due to rising prevalence and longer survival times (King's Fund, 2022). These statistics underscore the urgency of optimising rehabilitation pathways to improve mobility and independence post-stroke.

1.2 Stroke Rehabilitation and Current Assessment Practices

Rehabilitation is central to stroke care, with the aim of maximising recovery through neuroplasticity, task-specific practice, and compensatory strategies where necessary. The ability to regain walking function is strongly associated with quality of life, social participation, and reduced caregiver burden (Perry et al., 1995; Jørgensen et al., 1995). As such, the accurate assessment of gait is vital not only for monitoring progress but also for tailoring interventions.

1.2.1 Clinical outcome measures

In routine practice, gait and balance are often evaluated using standardised clinical tools. The Berg Balance Scale (BBS) provides a composite measure of postural control but is susceptible to ceiling effects in ambulatory stroke survivors (Blum & Korner-Bitensky, 2008). The Timed Up and Go (TUG) test evaluates functional mobility but does not discriminate between compensatory and restorative strategies (Podsiadlo & Richardson, 1991). The 10-Metre Walk Test (10MWT) and 6-Minute Walk Test (6MWT) are widely used to measure walking speed and endurance, with walking speed frequently described as the “sixth vital sign” due to its predictive value for function and survival (Middleton et al., 2015). Categorical measures such as the Functional Ambulation Categories (FAC) are commonly employed in discharge planning but offer limited mechanistic insight (Holden et al., 1991).

Although these measures are considered reliable and feasible in clinical environments, they provide no information about the underlying biomechanics or neuromuscular mechanisms of gait recovery (Kwakkel et al., 2019). For example, a patient may demonstrate improved walking speed through compensatory hip hiking rather than restored plantarflexor function. As such, outcome measures alone cannot distinguish between true recovery and compensatory adaptation (Balasubramanian et al., 2007). These limitations, and their implications for post-stroke gait evaluation, are explored further in Chapter 2, Section 2.4.1, which critically reviews clinical outcome measures in relation to biomechanical and neuromuscular markers.

1.3 Paretic Gait and Mechanistic Insights

Post-stroke hemiplegic gait is characterised by multiple deviations, including reduced step length, stance duration, symmetry and propulsion from the paretic limb (Olney & Richards, 1996). These abnormalities are underpinned by deficits in motor control; muscle activation and inter-joint coordination. Impaired plantarflexor function is a key limitation, as the gastrocnemius and soleus have been seen to generate ~85% of the forward propulsive force during forward propulsion (Neptune et al., 2001; Bowden et al., 2006).

Research into muscle synergies has revealed reduced ability to modulate activation patterns in stroke survivors, resulting in a loss of selective motor control (Clark et al., 2010; Cheung et al., 2012). This neuromuscular simplification contributes to inefficient gait strategies and limited adaptability. Understanding these mechanisms is essential for designing targeted interventions, yet in practice gait evaluation often remains superficial. This contrast between mechanistic understanding and clinical practice is developed in Chapter 2, Section 2.3, which reviews normal and paretic gait in detail.

1.4 Measuring Gait: From Laboratory to Clinic

1.4.1 Three-dimensional gait analysis (3DGA)

3DGA provides gold-standard kinematic and kinetic data and has advanced understanding of post-stroke gait abnormalities, including impaired gastrocnemius activation and altered timing of plantarflexor push-off (Den Otter et al., 2006; Clark et al., 2010). It's high cost, specialised expertise, and limited accessibility, however, constrain its role in routine rehabilitation (Carse et al., 2013). Moreover, 3DGA's ecological validity is limited, as laboratory settings do not reflect real-world walking (LeCursi et al., 2024). These challenges are further discussed in Chapter 2, Section 2.4.2, which compares 3DGA with observational gait analysis and emerging hybrid tools.

1.4.2 Shank-to-vertical angle (SVA)

To bridge the gap between laboratory precision and clinical feasibility, simpler biomechanical markers have been explored. The shank-to-vertical angle (SVA)—defined as the orientation of the tibia relative to vertical during stance—has emerged as a practical and adaptable measure (Owen, 2004). SVA directly influences knee stability, hip extension, and forward propulsion and can be measured using simple observational or inertial sensor-based methods (Kerkum et al., 2015).

Evidence suggesting an optimised SVA improves symmetry, efficiency, and stability in stroke survivors is beginning to emerge (Bregman et al., 2012; Harper et al., 2014; Kerr et al., 2020). Importantly, unlike propulsion which requires force plates or motion capture, SVA is both a measurable outcome and an adjustable intervention parameter. Through orthotic tuning and footwear modification, clinicians can directly manipulate SVA to improve gait

mechanics (LeCursi et al., 2024). The role of SVA in clinical decision making as both outcome measure and intervention parameter is critically reviewed in Chapter 2, Section 2.4.4.

1.5 Orthotic Practice in Stroke Rehabilitation

1.5.1 The role of ankle–foot orthoses (AFOs)

AFOs are widely prescribed to improve swing-phase clearance, stability, and efficiency (Tyson & Kent, 2013). National guidelines recommend their use, but their prominence in practice far exceeds the limits of their recommendations in policy documents such as NICE CG162, where they are only briefly mentioned (NICE, 2013). Orthotic evidence and debates are examined in Chapter 2, Section 2.5, which explores AFO design, tuning, and integration with rehabilitation.

1.5.2 Muscle inhibition debate

A long-standing concern is that AFOs may suppress muscle recruitment, particularly in the dorsiflexors and plantarflexors, by externally guaranteeing stability (Hesse et al., 1999; Lairamore et al., 2011). This has led to debate in the literature between neuroplasticity with muscle activation and the perceived safety and stability offered through rigid orthoses. However, longitudinal studies suggest that fears of long-term inhibition are overstated: Wearing et al. (2019) and Choo et al. (2021) found no evidence of progressive muscle atrophy when AFOs were integrated into active rehabilitation. This debate is discussed in depth in Chapter 2, Section 2.5.1–2.5.2, where evidence both supporting and refuting the inhibition hypothesis is critically reviewed.

1.5.3 Influence of design and tuning

AFO effectiveness depends on design. Articulated or spring-like designs can facilitate more natural kinematics and reduce metabolic cost, while rigid devices may restrict propulsion (Mulroy et al., 2010; Bregman et al., 2012; Mahmood et al., 2022). The SVA provides an objective framework for tuning orthoses, ensuring that alignment promotes knee stability and propulsion (Kerkum et al., 2015; Kerr et al., 2020). Poorly tuned devices, however, may inadvertently impede safe toe clearance (Burpee & Lewek, 2015). The mechanistic basis of tuning and the clinical relevance of poor alignment are addressed in Chapter 2, Section 2.5.3.

1.5.4 Orthoses in integrated rehabilitation

Recent evidence-based advances in neurorehabilitation practice emphasise high-intensity, task-specific, and engaging practice, supported by robotics, treadmill training, and gamification (Duncan et al., 2011; Maier et al., 2019; Mehrholz et al., 2020). AFOs should not be seen as stand-alone solutions but as enablers of participation in such training. Integrating orthotic tuning with EMG biofeedback and other mechanistic assessments provides a more holistic, individualised approach to gait rehabilitation. These interactions are expanded in Chapter 2, Section 2.5.5, which examines orthoses within broader adjunctive rehabilitation strategies.

1.6 Rationale and Thesis Focus

Taken together, the evidence highlights the need for clinically feasible yet mechanistically informative outcome measures in stroke rehabilitation. Functional scales remain indispensable but cannot distinguish recovery from

compensation; 3DGA offers insight but is impractical for routine use. SVA and EMG represent promising intermediate solutions, bridging biomechanics and clinical practice. Orthoses, particularly when tuned with reference to SVA, are a cornerstone of rehabilitation but require further integration with neuromuscular measures to ensure they promote recovery rather than compensation.

This thesis therefore focuses on the development and evaluation of wearable EMG and SVA as tools for assessing and optimising gait in stroke survivors, with particular attention to their integration into orthotic practice. In doing so, it seeks to bridge the gap between laboratory biomechanics and real-world rehabilitation. The following chapter (Chapter 2) expands upon these themes through a structured review of the literature, beginning with normal and paretic gait (Section 2.3), before considering gait measurement strategies (Section 2.4), orthotic interventions (Section 2.5), and a systematic review of recent rehabilitation trials (Section 2.6).

Chapter 2 – Literature Review

2.1 Introduction

As discussed through the Introduction Chapter 1, Stroke remains one of the leading causes of long-term disability worldwide, with gait impairment being among the most common and persistent consequences (Dobkin, 2005; Winstein et al., 2016). Post-stroke gait dysfunction is multifactorial, with biomechanical measurable changes resulting from a combination of neurological and musculoskeletal deficits. This has led to considerable research into gait rehabilitation, where the integration of biomechanical analysis and surface electromyography (EMG) has been particularly influential in identifying deficits that can guide therapeutic interventions.

This literature review has been intentionally structured to first examine the broader theoretical and clinical concepts underpinning the thesis before progressing to the focused systematic review presented in Section 2.6. Initial sections therefore explore gait biomechanics, stroke-related gait dysfunction, clinical gait assessment, orthotic intervention, and current movement analysis and measurement approaches in order to establish the wider clinical, biomechanical, and technological context surrounding the research programme. This staged approach was considered necessary due to the interdisciplinary nature of the thesis, which draws upon principles from rehabilitation science, orthotics, biomechanics, clinical movement analysis, wearable technology, and neuromuscular assessment.

The breadth of this preliminary review also reflects the evolving themes that emerged throughout the research process and the relative scarcity of literature directly examining the combined application of orthotic intervention and electromyography (EMG)-informed gait assessment within stroke

rehabilitation. While substantial literature exists relating independently to stroke gait rehabilitation, orthotic management, and EMG analysis, there remains comparatively limited evidence exploring their integration within clinically applicable assessment systems and real-world orthotic practice. Previous authors have also identified wider limitations and inconsistencies within orthotic rehabilitation literature, including variability in reporting standards, participant characterisation, intervention description, and outcome measurement, which further restrict direct comparison across studies (Golding-Day et al., 2024).

Consequently, a broader narrative review was required initially to capture the range of interconnected concepts relevant to the thesis aims before undertaking the more focused systematic literature review presented in section 2.6. This structure allowed both contextual exploration of the wider evidence base and critical examination of the specific knowledge gaps that informed the subsequent study design and methodological development.

2.2 Normal Gait and Biomechanics of Walking

Normal gait is an efficient, symmetrical process that balances three objectives: maintaining stability, generating propulsion, and adapting to environmental variability (Winter, 1991; Perry & Burnfield, 2010). These objectives are achieved through tightly coordinated neuromuscular patterns, shaped by both cortical control and spinal networks, and modulated by sensory feedback (Grillner, 2006; Dietz, 2003).

A defining feature of normal gait is its efficiency. The body conserves energy by transferring potential and kinetic energy between steps in an inverted pendulum-like fashion (Kuo, 2007). Within this process, the plantarflexors—particularly the gastrocnemius–soleus complex—play a disproportionately

important role, contributing over half of the positive work of walking and providing critical forward propulsion (Neptune et al., 2001; Farris & Sawicki, 2012). Their contribution to gait economy makes them a focal point in rehabilitation research, since disruption of plantarflexor function is strongly linked to post-stroke walking inefficiency (see Section 2.3).

Healthy gait is also characterised by symmetry and predictability. Step length, timing, and limb loading are balanced between sides, producing smooth forward progression with minimal mediolateral sway. Ground reaction forces follow a distinctive double-peaked pattern reflecting controlled braking and propulsion. These features are not simply aesthetic but are markers of neuromuscular integrity; their disruption is associated with instability, elevated energy cost, and reduced community ambulation (Perry & Burnfield, 2010).

The significance of these normative features lies not in their description but in what their absence reveals. Symmetry and coordinated segmental motion provide a lens through which the consequences of stroke can be evaluated. In the next section (2.3), the breakdown of these mechanisms in stroke survivors is examined, highlighting how impairments in neural control and muscle activation manifest as the distinctive characteristics of paretic gait.

2.3 Stroke and hemiplegic Gait

Stroke disrupts the coordinated neural control required for efficient gait, producing characteristic patterns of asymmetry, reduced propulsion, and compensatory movement strategies. The consequences are not uniform: they reflect the site and severity of brain injury, the balance between corticospinal and brainstem pathways, and the survivor's capacity for neuroplastic adaptation (Krakauer & Carmichael, 2017; Kwakkel et al., 2019).

Nevertheless, certain gait features are consistently observed in the post-stroke population, giving rise to the term hemiplegic gait.

A hallmark impairment is the loss of distal motor control. Weakness and delayed activation of the gastrocnemius–soleus complex diminish the forward propulsion (propulsive impulse) generated in late stance (terminal stance–pre-swing), forcing reliance on hip hiking or circumduction to achieve limb advancement or swing (Lamontagne et al., 2000; Bowden et al., 2006). Similarly, impaired tibialis anterior function produces foot drop during swing, compromising toe clearance and increasing fall risk. These deficits are often compounded by spastic co-contraction, where inappropriate timing of the plantarflexor or quadriceps muscles further restricts efficient movement (Clark et al., 2010).

The combined result is reduced forward propulsion, slower walking speeds, and greater metabolic cost compared to healthy gait (Farris & Sawicki, 2012; Awad et al., 2016). In healthy walking, the ankle plantarflexors contribute approximately 41–46% of the limb's total positive joint power across walking speeds (Farris & Sawicki, 2012), highlighting their central role in generating forward propulsion.

Following stroke, diminished plantarflexor output reduces this propulsive capacity, contributing to slower self-selected walking speeds, for example, Awad et al. (2016) reported a mean comfortable speed of 0.71 ± 0.05 m/s in individuals with chronic hemiparesis, compared to about 1.25 m/s in healthy adults. This impaired propulsion also increases the energetic cost of transport, with post-stroke individuals consuming roughly 61% more oxygen per meter than healthy controls (0.29 ± 0.02 vs. 0.18 ± 0.002 mL O₂·kg⁻¹·m⁻¹, Awad et al., 2016). Interventions that restore paretic propulsion, can reduce this metabolic cost by up to 24% at comfortable speeds (Awad et al., 2016).

Another defining feature of hemiplegic gait is asymmetry. Hemiplegic survivors frequently display shorter step length and reduced stance time on the affected limb, leading to instability and compensatory weight transfer onto the non-paretic side (Patterson et al., 2008). This asymmetry is not only a biomechanical issue but also a clinical predictor of long-term walking capacity and community participation (Perry et al., 1995). Importantly, improvements in walking speed alone may not reflect improved symmetry, as faster gait can be achieved through exaggerated compensations (Balasubramanian et al., 2007).

From a neuromechanical perspective, stroke reduces the modularity of muscle activation patterns, with multiple pre-stroke synergies merging into fewer, less flexible modules (Clark et al., 2010; Cheung et al., 2012). This reduced adaptability constrains responses to task demands, such as navigating uneven terrain or increasing speed, and contributes to falls and fatigue. These findings illustrate that paretic gait is not merely a slower version of normal gait but represents a qualitatively different motor strategy.

The clinical implications of these abnormalities are profound. Although many survivors regain some degree of walking independence, typically reflected by the ability to walk short household distances, their endurance and stability remain markedly reduced. Community-based cohort studies have shown that fewer than 15–20 % of stroke survivors achieve truly independent community ambulation at one-year post-stroke, commonly defined as walking at least 500 m without physical assistance (Lord et al., 2004; Michael et al., 2005). Even among those who do, gait speed and endurance remain well below normative values—average comfortable walking speeds of 0.4–0.8 m/s contrast sharply with the 1.2–1.4 m/s typical of healthy adults (Perry et al., 1995; Tilson et al., 2010).

These persistent limitations have substantial functional consequences, including reduced participation in outdoor and social activities, heightened fall risk, and continued reliance on assistive devices or caregiver support (Shumway-Cook et al., 2002; Patterson et al., 2008).

Traditional outcome measures capture *whether* individuals regain walking function but not *how* it is achieved - an important distinction when compensatory strategies dominate (see Section 2.4). This gap underpins the rationale for incorporating mechanistic markers such as EMG and SVA into rehabilitation assessment, as explored later in this chapter.

2.4 Measuring Hemiplegic Gait

The accurate measurement of post-stroke gait is essential for evaluating rehabilitation outcomes and guiding clinical decision-making. Yet there remains a persistent tension between functional outcome measures, which are easy to administer but limited in mechanistic insight, and instrumented biomechanical measures, which offer precision but are often impractical in routine care. This section reviews the principal approaches to gait assessment in stroke, highlighting their strengths, limitations, and potential integration.

2.4.1 Clinical outcome measures

Standardized clinical scales remain the most widely used method of gait assessment due to their ease of administration, high reliability, and broad applicability across clinical settings. Tools such as the Berg Balance Scale (BBS) and the Timed Up and Go (TUG) provide valuable information on postural stability and functional mobility. The BBS demonstrates excellent inter- and intra-rater reliability in post-stroke populations (intraclass correlation coefficient [ICC] = 0.98–0.99), with a minimal detectable change (MDC) of

approximately 2.7–7 points, depending on stroke chronicity (Berg et al., 1992; Blum & Korner-Bitensky, 2008). Similarly, the TUG shows excellent test–retest reliability ($ICC \approx 0.98$) and strong correlations with balance and gait performance measures such as the BBS ($r = -0.81$) and walking speed ($r = -0.61$) (Podsiadlo & Richardson, 1991; Flansbjer et al., 2005).

Walking speed, often measured by the 10-Metre Walk Test (10 MWT), is considered a “vital sign” of post-stroke recovery (Bohannon, 1997; Perry et al., 1995). The 10 MWT has shown excellent reliability ($ICC = 0.76–0.98$) and sensitivity to clinically meaningful change, with a standard error of measurement (SEM) around 0.36 m/s and an MDC of approximately 1.0 m/s in acute stroke (Tilson et al., 2010). Endurance is commonly evaluated using the 6-Minute Walk Test (6 MWT), which also demonstrates excellent test–retest reliability ($ICC = 0.83–0.99$) and an MDC ranging from 25–75 m, depending on patient status (Flansbjer et al., 2005; Fulk et al., 2008). Finally, the Functional Ambulation Categories (FAC), a six-point ordinal scale of ambulation dependence (Holden et al., 1986, 1991), provides a quick categorical assessment of walking ability with high test–retest ($\kappa = 0.95$) and inter-rater reliability ($\kappa = 0.91$), and strong concurrent validity with both walking speed and endurance (Mehrholz et al., 2007).

While reliable and pragmatic, these measures share key limitations. They primarily capture *performance* rather than *mechanism*, offering little insight into whether gains reflect neurophysiological recovery or compensatory strategies (Balasubramanian et al., 2007; Kwakkel et al., 2019). Many also suffer from ceiling effects, particularly as patients transition to community ambulation, reducing sensitivity to clinically meaningful change (Blum & Korner-Bitensky, 2008). As a result, although these measures remain indispensable for service evaluation and communication across disciplines, they cannot fully explain the dynamics of paretic gait.

2.4.2 Three-dimensional and observational gait analysis

At the opposite end of the spectrum, three-dimensional gait analysis (3DGA) provides a gold-standard, laboratory-based assessment of joint kinematics, kinetics, and spatiotemporal parameters. By combining marker-based motion capture, synchronised force plates, and often EMG, 3DGA can identify impairments such as delayed gastrocnemius activation, abnormal quadriceps timing, or reduced plantarflexor push-off (Den Otter et al., 2006; Clark et al., 2010). It also provides insight into neuromuscular organisation: for example, altered or merged synergies have been linked to cortical damage (Cheung et al., 2012), and combined kinematic–kinetic–synergy models predict impairment more effectively than clinical scales alone (Barroso et al., 2017).

Yet, three-dimensional gait analysis (3DGA) remains constrained by practical feasibility. A single session can require 1.5–3 hours, encompassing marker placement, calibration, multiple walking trials, and post-processing, which demands extensive technical expertise and specialized laboratory infrastructure (Kelly et al., 2016). Moreover, the process can be physically and cognitively demanding, with up to 40–50 % of stroke survivors unable to complete full protocols due to fatigue, limited endurance, or attentional deficits (Kelly et al., 2016; National Institute for Health Research [NIHR], 2018). National guidance emphasises multidisciplinary, needs-led care but does not identify 3DGA as a core service (NICE, 2023), reflecting both its cost and limited clinical availability. In the UK, accredited laboratories are clustered around major centres, restricting access for many patients. Consequently, 3DGA is generally reserved for complex cases such as spasticity management, orthotic prescription, or surgical planning (Carse et al., 2013).

By contrast, observational gait analysis (OGA) is widely used in clinical practice. Clinicians rely on visual inspection, sometimes supported by structured scales such as the Wisconsin Gait Scale, to identify deviations

including circumduction, hip hiking, or foot drop (Toro et al., 2007; Rodriguez et al., 2019). While OGA lacks the temporal and spatial precision of 3DGA, its strengths lie in feasibility: it is repeatable, requires minimal equipment, and can be applied at the bedside or in community settings.

The contrast between 3DGA and OGA reflects a broader methodological divide: laboratory methods deliver precision but lack scalability, while clinical tools offer practicality but risk oversimplification.

2.4.3 Emerging hybrid approaches and service models

Recent developments aim to bridge the gap between laboratory precision and clinical feasibility. Wearable inertial sensors and portable EMG systems now enable collection of spatiotemporal and neuromuscular data in real-world environments (Picerno, 2017). These tools align with mechanistic priorities identified in the research literature: paretic propulsion (Bowden et al., 2006; Awad et al., 2016), trailing limb angle (Peterson et al., 2010; Hsiao et al., 2015), and muscle synergy modularity (Clark et al., 2010; Cheung et al., 2012; Barroso et al., 2017) are all established determinants of functional recovery. If validated for clinical use, wearable systems could embed these parameters into day-to-day rehabilitation, moving beyond descriptive observation to mechanism-informed practice.

Still, validation and integration remain incomplete. Many stroke services lack the infrastructure to adopt hybrid systems, and methodological consistency (e.g., sensor placement, EMG signal processing) remains a barrier to widespread uptake (Baudet et al., 2014). Until such challenges are resolved, hybrid tools remain a promising but transitional solution.

2.4.4 Shank-to-vertical angle (SVA) as a pragmatic marker

The shank-to-vertical angle (SVA) has emerged as one candidate for bridging this divide. Defined as the orientation of the tibia relative to vertical during stance, SVA directly influences lower-limb kinetics, including knee stability, hip extension, and forward propulsion (Owen, 2004; Mulroy et al., 2010). Unlike propulsion or trailing-limb angle, which require force plates or motion capture, SVA can be measured through inertial sensors or even visual observation, making it highly adaptable to clinical practice (Kerkum et al., 2015).

Evidence suggests that optimising SVA can improve gait symmetry, efficiency, and stability in stroke survivors (Bregman et al., 2012; Harper et al., 2014; Kerr et al., 2020). Importantly, SVA is not only an outcome measure but also an intervention parameter: it can be systematically adjusted through orthotic and footwear tuning (LeCursi et al., 2024).

This dual role positions SVA as a practical tool for bridging the gap between clinical pragmatism and biomechanical validity. However, variability across individuals and measurement methods remains a challenge, and relatively few large-scale trials have validated SVA as a primary outcome.

2.4.5 Electromyography (EMG) as a mechanistic marker

Electromyography (EMG) provides a mechanistic layer of analysis within gait assessment by directly quantifying neuromuscular activation patterns. Whereas kinematic and kinetic analyses describe movement outcomes, EMG identifies the neural control strategies responsible for generating those movements. This makes EMG particularly valuable in post-stroke populations, where apparent improvements in gait may stem from compensatory rather than restorative mechanisms. Across studies, EMG has consistently highlighted altered activation timing, excessive co-contraction, and reduced

modular organisation of muscle synergies (Clark et al., 2010; Cheung et al., 2012), establishing it as a critical tool for interpreting the neuromechanical origins of gait impairment.

Historically, EMG was restricted to specialist research environments because of the technical complexity of electrode preparation, noise reduction, and calibration. Early systems relied on wired electrodes and cumbersome amplifiers, resulting in protracted data collection protocols and limited clinical feasibility (Bonato, 2010). These practical barriers confined EMG to small, controlled laboratory samples, and the resulting data, though rich, were rarely incorporated into clinical decision-making.

The emergence of wireless, textile-based, and high-density EMG (HD-EMG) technologies has transformed both feasibility and resolution. Wireless systems have streamlined setup and improved signal quality during overground walking, while textile electrodes and dry sensors have made longer-duration and community-based monitoring possible (Finni et al., 2007; Santuz et al., 2020). HD-EMG arrays now permit spatial mapping of motor-unit recruitment, revealing previously inaccessible aspects of neural drive, coordination, and fatigue (Hug et al., 2021). Collectively, these advances have repositioned EMG from a research-only modality toward a credible component of modern gait analysis, with increasing potential for clinical integration.

Within gait laboratories, EMG is most powerful when synchronised with kinematic and kinetic data. This integration allows assessment of the temporal coordination between muscle activation, joint kinetics, and ground reaction forces, providing causal links between neural control and biomechanical output. EMG-driven musculoskeletal models now permit estimation of muscle-specific contributions to joint moments and power, enabling researchers to quantify neuromuscular efficiency and the relative contribution of different muscle groups to propulsion and support.

Furthermore, the application of synergy analysis, using techniques such as non-negative matrix factorisation, has enabled dimensional reduction of complex EMG datasets, offering a compact representation of neuromuscular coordination. Post-stroke reductions in synergy number and independence have been widely interpreted as evidence of impaired neural modularity (Clark et al., 2010; Cheung et al., 2012), although methodological sensitivity to filtering, factor extraction, and normalisation remains a key limitation (Santuz et al., 2020).

EMG's clinical relevance extends beyond diagnostic insight to therapeutic monitoring and feedback. When used in real time, EMG biofeedback can support re-education of selective muscle activation, enhance motor awareness, and reinforce gait symmetry (Skvortsov et al., 2021). Such approaches are increasingly used alongside robotic and functional electrical stimulation interventions to guide neuromuscular retraining. Despite this promise, the evidence base remains fragmented: protocols vary widely in electrode configuration, threshold criteria, and training dosage, limiting generalisability. Moreover, the cognitive and attentional demands of biofeedback can be challenging for individuals with post-stroke fatigue or cognitive deficits, constraining uptake in clinical practice.

While EMG provides unique mechanistic insight, several constraints continue to limit its routine clinical adoption. Signal quality depends heavily on electrode placement, cross-talk suppression, and normalisation methods, all of which can vary across laboratories and influence comparability (Hug et al., 2021). Interpretation also requires a level of technical expertise that exceeds most clinical settings, particularly when analysing multi-muscle or synergy-level data. Additionally, EMG amplitude does not translate linearly to muscle force due to differences in muscle fibre composition, electrode geometry, and subcutaneous tissue thickness, factors that are often accentuated in

pathological populations. These methodological sensitivities mean that EMG findings, although mechanistically informative, are seldom used for clinical decision-making in isolation.

From a conceptual standpoint, EMG studies have often remained descriptive, focusing on the presence of abnormal activation patterns rather than quantifying how these patterns affect biomechanical or metabolic outcomes. Bridging this gap requires integrative modelling approaches that combine EMG with inverse dynamics, musculoskeletal simulation, or energy cost estimation, allowing neural activation to be directly linked with gait efficiency and recovery potential. The increasing convergence of wearable EMG, motion capture, and machine learning offers a promising pathway toward such integration.

2.4.6 Summary of Measuring Hemiplegic Gait

Measuring paretic gait requires balancing feasibility with mechanistic insight. Clinical scales remain indispensable but limited; 3DGA provides precision but is impractical for widespread use; OGA is feasible but lacks mechanistic depth. Hybrid systems, SVA, and EMG offer promising middle-ground approaches that may embed mechanistic insights into clinical practice. The integration of these measures represents a promising direction for rehabilitation, particularly in orthotic management, which is discussed in the next section (2.5).

2.5 Orthotic Practice in Stroke Rehabilitation

Ankle–foot orthoses (AFOs) remain one of the most widely prescribed interventions for hemiplegic stroke survivors with gait impairments. Their purpose is to stabilise the ankle, enhance toe clearance, improve knee control, and enable safer ambulation (Tyson & Kent, 2013). Despite this widespread

use, their role within stroke rehabilitation is far from uncontested. AFOs sit at the intersection of clinical pragmatism and mechanistic debate: they provide immediate functional benefits but raise concerns about long-term influence on neuromuscular recovery.

This tension is reflected in clinical guidelines. While the NICE Stroke Rehabilitation Pathway advocates a multidisciplinary approach to recovery (NICE, 2019b), orthoses receive only brief mention in the Stroke Rehabilitation in Adults guideline (CG162, section 1.9.27), which advises clinicians to “consider ankle–foot orthoses for people who have difficulty with swing-phase foot clearance” (NICE, 2013). In practice, however, AFOs are routinely used to address foot drop, improve stability, and reduce energy cost. This discrepancy between guideline emphasis and real-world practice underscores the need for robust, mechanistically informed evidence.

2.5.1 Theoretical concerns about muscle inhibition

One of the more frequently voiced concerns among rehabilitation clinicians is that use of ankle-foot orthoses (AFOs) may produce muscle inhibition rather than functional recovery. The historical, theoretical, basis for this critique is straightforward. Rigid AFOs, by mechanically stabilising the ankle and ensuring toe-clearance during swing, reduce the net demand placed on the ankle dorsiflexors and plantarflexors during gait. In turn, this predicted reduction in activation is argued to undermine muscle recruitment, neuromuscular drive and the very basis of use-dependent neuroplastic change (Kirker, 2022). For example, Hesse et al. (1999) demonstrated that hemiparetic patients walking with a double-stopped rigid AFO exhibited reduced Tibialis Anterior (TA) activation compared with walking without the device - reinforcing concerns that AFOs might promote disuse of key ankle musculature. Lairamore et al. (2011) similarly reported diminished TA EMG in adults post-stroke during AFO-assisted walking. These findings underpin a

dominant physiotherapy narrative: that full benefit from gait rehabilitation requires active muscle engagement, and that compensatory devices risk locking in passive patterns rather than enabling recovery.

From this viewpoint, an AFO may inadvertently reinforce compensatory strategies rather than restore true motor recovery. If the mechanical role of the ankle is assumed by the device, the patient may be less inclined to recruit the TA during swing; likewise, if the device constrains ankle plantarflexion, the gastro-soleus complex may fail to engage in propulsive work. This is not simply a debate about stability versus mobility: it reflects deeper professional assumptions: physiotherapists emphasise neuro-plasticity, motor learning and muscle activation, whereas orthotists may historically tend towards emphasising stability, safety and immediate functional gain. The critical question becomes: does the AFO support *recovery* (active muscle work) or simply *compensation* (mechanical assistance)?

2.5.2 Evidence challenging the inhibition hypothesis

Conversely however, recent longitudinal and systematic work have begun to challenge the idea that AFOs necessarily inhibit muscle activation or delay rehabilitation. For example, Nikamp et al. (2019) randomised 26 post-stroke patients to early or delayed AFO provision: while TA activation during swing was significantly lower when walking *with* the device within a single session ($p = .041$), over the 26-week follow-up walking *without* an AFO showed no significant change in TA activation (within-groups $p = .420/0.282$; between-groups $p = .987$), suggesting no long-term deconditioning effect. Concurrently, a systematic review of 20 studies (Kenworthy et al., 2024) found that seven longitudinal studies did not report adverse long-term EMG changes with AFO use. More recently, Liu et al. (2024) found that in 20 sub-acute stroke patients, walking *with* a solid AFO resulted in increased TA sMMG signals ($p < .05$) and reduced TA–MG (medial gastrocnemius) co-contraction (from ~50 % to ~22

%), consistent with more “normalised” ankle muscle activation. Collectively, these data suggest that AFOs, when well designed and integrated within an active rehabilitation programme, may support rather than suppress muscle function.

However, the evidence is not unequivocal, and several caveats apply. First, the heterogeneity of orthosis designs (rigid vs articulated, fixed vs dynamic, posterior-leaf vs spring-assist) means that one cannot generalise broad statements about “AFOs” without specification: Kenworthy et al. (2024) emphasise that “the heterogeneity of AFO designs prevents broad statements related to which orthoses optimise muscle activity.” Second, many studies focus on EMG amplitude as the outcome, which is a proxy for activation rather than muscle strength, morphology or neuroplastic change. Third, many trials lack a design arm with *no* orthosis over the same follow-up period, raising potential confounds (walking speed, volume, training dose). Fourth, device design seems critical: older work by Mulroy et al. (2010) showed that AFOs that permit some ankle motion resulted in more favourable knee kinematics and muscle activation compared with rigid devices.

Taken together, the evidence points to an evolution in the debate: it is no longer tenable to argue that AFOs *automatically* inhibit muscles, but it remains valid (and clinically important) to ask under what conditions they might, and how device design and training context might mitigate risks.

2.5.3 Influence of AFO design and tuning

Not all AFOs act in the same way. Design characteristics—including rigidity, articulation, and energy-storing features—substantially influence their effect on gait. Articulated AFOs may permit controlled ankle motion and facilitate gastrocnemius recruitment, while energy-storing designs reduce metabolic cost and promote more natural kinematics (Bregman et al., 2012; Harper et

al., 2014; Vistamehr et al., 2024). Conversely, rigid plantarflexion-stop designs provide maximal stability but may excessively constrain ankle dynamics.

The importance of tuning further complicates interpretation. AFOs should not be viewed in isolation but as part of an AFO–Footwear Combination (AFO-FC). Adjustments to heel height, rocker sole design, and footplate stiffness can substantially alter shank inclination, knee moments, and hip extension. The shank-to-vertical angle (SVA) has emerged as a practical parameter to guide this tuning, providing an objective target for optimising gait (Owen, 2004; Kerkum et al., 2015; Kerr et al., 2020; LeCursi et al., 2024). Poorly tuned devices may not only reduce efficiency but risk blocking propulsion. Burpee and Lewek (2015) observed that the angular difference between a “trip” and a “clear” step could be as little as 10°, emphasising how sensitive gait is to alignment. Their findings highlight the importance of careful adjustment, as misaligned orthoses or fixed-timing FES devices could inadvertently impede safe toe clearance and paretic propulsion.

2.5.4 Orthoses compared with alternative interventions

The debate around AFOs often intersects with comparisons to functional electrical stimulation (FES) for foot drop. Randomised controlled trials consistently show that both interventions improve walking speed and endurance, with no clear superiority of one over the other (Kluding et al., 2013; Everaert et al., 2013; Bethoux et al., 2014). Patients sometimes express preference for FES due to comfort and aesthetics, while clinicians value the reliability and immediacy of AFOs. Importantly, both can be effective when integrated into therapy, and the choice is best framed around patient-specific needs and context rather than competition between modalities.

2.5.5 Integration with adjunctive rehabilitation strategies

Recent advances in rehabilitation emphasise intensity, task-specific practice, and engagement, principles that are increasingly supported by technologies such as robotics, treadmill training, virtual reality, and gamification (Duncan et al., 2011; Mehrholz et al., 2020; Maier et al., 2019). Within these contexts, AFOs may function less as passive supports and more as enablers of training, providing the stability required for survivors to engage in demanding, high-dose practice.

Yet, there remains a research–practice gap. Many propulsion-targeted gait training studies exclude AFO users altogether (e.g., Burpee & Lewek, 2015; Jonkers et al., 2009), creating results that are difficult to generalise to everyday clinical populations. Similarly, Pomeroy et al. (2012) demonstrated that early casting and orthotic provision were feasible, but their functional impact varied, raising questions about timing, device type, and integration with physiotherapy. These findings reinforce the need for trials that explicitly test combined therapy-plus-orthosis strategies, with outcomes extending beyond speed to propulsion, endurance, and real-world activity.

2.5.6 Summary of Orthotic Intervention in Stroke Rehabilitation

Across the preceding sections, the literature reveals both the clinical ubiquity and evidential fragility of AFOs within post-stroke rehabilitation. Despite decades of use, methodological inconsistencies have limited consensus on their long-term neuromuscular effects. Studies vary widely in orthosis design, participant chronicity, gait task selection, and outcome definition, rendering inter-study comparison problematic. This variability partly explains the persistent polarity between professional groups: physiotherapists, often grounded in neuroplastic models of recovery, tend to view AFOs with caution;

orthotists and rehabilitation engineers, guided by biomechanical optimisation, emphasise stability and energy efficiency (Kirker, 2022).

Methodologically, early evidence supporting the muscle-inhibition hypothesis drew heavily on *acute, single-session EMG studies* (e.g., Hesse et al., 1999; Lairamore et al., 2011). These designs quantified immediate reductions in tibialis anterior amplitude when rigid AFOs were introduced but rarely controlled for confounding factors such as walking speed, surface variability, or learning effects. Without longitudinal follow-up or concurrent morphological assessment, such findings cannot distinguish short-term mechanical unloading from genuine neurophysiological suppression. Small samples (often $n < 10$) and inconsistent normalisation methods further undermine external validity. The frequent absence of concurrent kinetic or kinematic data also precludes linking observed EMG changes to functional outcomes such as propulsion or gait efficiency.

In contrast, more recent longitudinal and comparative studies have improved methodological rigour through randomisation, repeated-measures designs, and multi-modal outcome sets. For example, Nikamp et al. (2019) and Wearing et al. (2018, 2019) incorporated follow-up intervals up to 26 weeks, demonstrating that initial EMG reductions did not translate into progressive muscle decline. However, these trials still suffer from moderate sample sizes and lack of blinding. Meta-analytic efforts such as Kenworthy et al. (2024) highlight this heterogeneity explicitly, concluding that “evidence of long-term inhibition is absent, but confidence in the estimate remains low due to design variability” (p. 8). Such caveats illustrate a central problem in the AFO literature: methodological noise has often been mistaken for mechanistic truth.

Design-specific investigations provide richer mechanistic insight but also underscore the importance of orthosis tuning and context. Studies comparing articulated or spring-like devices with rigid configurations (Mulroy et al., 2010;

Bregman et al., 2012; Harper et al., 2014) demonstrate that limited ankle motion and energy-storing elements can preserve or even enhance gastrocnemius activity and reduce metabolic cost by approximately 10 %. Yet these trials frequently rely on prototype devices and laboratory protocols that do not mirror clinical practice.

Conversely, pragmatic clinical audits reveal that most prescriptions remain fixed-angle or prefabricated designs (Mahmood et al., 2022), suggesting a translational lag between experimental optimisation and real-world provision. The failure to report tuning parameters, such as heel height, or footwear configuration, represents a recurrent methodological omission that constrains reproducibility and clinical applicability.

Another limitation lies in outcome selection bias. Many trials privilege walking speed or endurance as primary outcomes, despite these metrics' inability to discriminate between compensatory and restorative mechanisms (Balasubramanian et al., 2007). Few studies concurrently measure propulsion, symmetry, or muscle synergy modularity, variables that better capture the quality of movement. Even fewer embed patient-reported measures of stability or fatigue, leaving the lived functional impact of orthotic intervention under-represented. Consequently, the evidence base remains functionally descriptive rather than mechanistically explanatory.

Synthesising across these methodological trends, several critical observations emerge. First, the long-standing dichotomy between *inhibition* and *facilitation* is oversimplified: EMG amplitude alone cannot adjudicate neuroplastic outcomes without corroborating strength or morphology data. Second, orthotic design and alignment profoundly moderate neuromuscular responses, rendering the term “AFO effect” analytically meaningless without specification. Third, rehabilitation context, particularly the intensity of concurrent physiotherapy, appears as influential as the device itself; studies integrating

active gait training with orthotic use consistently report superior outcomes (Choo et al., 2021). Finally, the methodological field suffers from a lack of integrated, multimodal metrics that connect neural activation (EMG) to functional performance (kinematics, kinetics, endurance).

Collectively, these insights point to a research landscape characterised less by contradiction than by methodological fragmentation. The apparent disagreements between professions and studies may stem more from design heterogeneity than from true mechanistic opposition. Resolving this requires systematic, transparent synthesis of intervention studies using consistent inclusion criteria, explicit reporting of device parameters, and outcomes that span both function and mechanism.

This methodological gap provides the rationale for the systematic literature review that follows. By applying structured PICO-based screening and critical appraisal frameworks, the subsequent review aims to clarify the extent to which current post-stroke gait interventions, orthotic or otherwise, capture the neuromechanical underpinnings of recovery rather than merely recording functional compensation.

2.6 Systematic Literature Review

2.6.1 Introduction

The preceding sections (2.2–2.5) have traced a trajectory from normal gait mechanics, through the altered neuromechanical profile of post-stroke paretic gait, to the diverse tools and interventions available for its assessment and rehabilitation. This synthesis established two critical points. First, although clinical outcome measures such as the 10-Metre Walk Test (10MWT) and Berg Balance Scale (BBS) remain indispensable in practice, they lack the granularity to distinguish compensation from true recovery. Second, advanced methods such as three-dimensional gait analysis (3DGA) and electromyography (EMG) provide mechanistic insight, yet they are impractical for routine use. Emerging approaches such as the shank-to-vertical angle (SVA) and wearable EMG devices appear promising but require validation before they can bridge the translational gap between laboratory precision and clinical feasibility.

A parallel tension surrounds orthotic interventions. While ankle–foot orthoses (AFOs) remain the mainstay of post-stroke walking rehabilitation, the evidence base is complicated by inconsistent study designs, heterogeneous outcome measures, and frequent exclusion of orthotic users from clinical trials. This prevents clear understanding of how orthotic tuning (particularly through SVA alignment) interacts with neuroplastic recovery, propulsion, and long-term functional outcomes.

Taken together, these gaps highlight the need for a systematic evaluation of the intervention literature. The aim of this review is therefore to critically examine post-stroke gait rehabilitation interventions, with specific attention to the outcome measures employed. This focus is not limited to functional

performance (e.g., walking speed), but extends to mechanistic markers of propulsion, synergy modularity, and limb alignment. In doing so, the review seeks to answer a central question: to what extent do current intervention studies capture the neuromechanical underpinnings of recovery, rather than simply recording compensatory improvements in function?

The following sections present the methods, findings, and critical interpretation of this systematic review.

2.6.2 Search Protocol

A systematic search strategy was developed to align with PRISMA (Preferred Reporting Items for Systematic Reviews and Meta-Analyses) standards. Searches were performed across MEDLINE, CINAHL, Embase, Web of Science, and the Cochrane Library with supplementary hand-searching of reference lists from included articles and relevant reviews. The search strategy was designed to identify relevant literature examining lower-limb muscle activity during gait following stroke. Each database strategy combined controlled vocabulary (e.g., MeSH, Emtree, and CINAHL Headings) with corresponding free-text terms in titles and abstracts to maximise sensitivity. Search terms encompassed variations of *stroke*, *gait* or *walking*, and *electromyography (EMG)*, while studies primarily investigating the upper limb were excluded through the use of specific NOT terms. (see Appendix A for full search strategies for each database).

The review adopted a PICO (Population, Intervention, Comparator, Outcome) framework to structure both the search and the subsequent data extraction. This decision reflects both methodological rigor and the clinical relevance of framing evidence in terms of patient population, targeted intervention, comparator conditions, and the specific outcomes measured. Unlike purely narrative syntheses, the PICO approach enables structured comparisons

across heterogeneous interventions, while also foregrounding the limitations of outcome selection.

2.6.2.1 Search Limits and Filters

The database searches were restricted to human studies in adult populations. Only articles published in peer-reviewed journals were included to ensure methodological quality and reproducibility. Filters were applied to exclude paediatric samples, non-English language studies, and non-empirical publications (such as editorials, commentaries, or study protocols without data). Publication date limits were imposed from January 2004 to December 2024. Where unclear or incomplete data were encountered in published articles, individual follow-up searches of supplementary materials and trial registries were conducted. If outcome data remained unavailable, the study was excluded to avoid compromising the integrity of extracted results.

2.6.3 Inclusion and Exclusion Criteria

Inclusion and exclusion criteria were defined a priori to ensure transparency and reproducibility.

2.6.3.1 Inclusion criteria:

- Population: Adults post-stroke (acute, sub-acute, or chronic phase).
- Intervention: Any gait-focused rehabilitation, including orthotic devices (AFOs), functional electrical stimulation (FES), treadmill or robotic training, propulsion-targeted protocols, and wearable/technological adjuncts.
- Comparator: Usual care, no intervention, or alternative gait interventions.

- Outcomes: Studies reporting gait outcomes, including spatiotemporal measures (e.g., speed, step length), biomechanical variables (e.g., propulsion, trailing limb angle, SVA), or neuromuscular indices (e.g., EMG, synergy analysis).
- Study type: Randomised controlled trials (RCTs), quasi-experimental studies, prospective cohorts, and controlled observational designs.

2.6.3.2 Exclusion criteria:

- Non-stroke neurological populations (unless stroke subgroup data were reported separately).
- Paediatric populations.
- Studies lacking quantitative gait outcomes.
- Case studies, conference abstracts, or protocols without data.
- Non-English language unless translation was feasible.
- These criteria were deliberately inclusive of a range of study designs, reflecting the reality that high-quality RCTs remain scarce in this domain. At the same time, exclusions avoided dilution with studies unable to inform mechanistic or functional gait outcomes.

2.6.3.3 Methodological Quality Appraisal

To reduce the risk of bias within the evidence base, methodological quality of included studies was appraised using the Cochrane Risk of Bias tool (RoB 2) (Higgins et al., 2022) for randomised controlled trials and the ROBINS-I (Sterne et al., 2016) framework for non-randomised and quasi-experimental studies. These instruments evaluate potential bias across key domains, including randomisation and allocation, deviations from intended interventions, outcome measurement, and selective reporting.

In this review, all assessments were conducted by a single reviewer (the author). While this introduces the possibility of bias in the appraisal process, it also ensured consistency of judgement across the full set of included studies. To mitigate bias, all decisions were guided by the published RoB 2 and ROBINS-I criteria, and where ambiguities arose, study protocols and supplementary materials were consulted to clarify methods and outcomes (Higgins et al., 2022; Sterne et al., 2016).

This approach reflects the reality of a heterogeneous evidence base in stroke rehabilitation, where exclusion of lower-quality studies risks narrowing the scope of analysis, but explicit acknowledgement of methodological weaknesses remains essential for critical appraisal.

2.6.4 Screening and PRISMA Flow Critical Overview

Following the PRISMA screening process and application of Inclusion and Exclusion criteria, (Section 2.6.2 and 2.6.3), a total of 83 studies were identified as meeting all inclusion criteria and were retained for full analysis. These studies are detailed in Appendix A, which presents the extracted data in a PICO (Population, Intervention, Comparison, Outcome) format. The PICO framework was adopted to ensure structured and transparent reporting, aligning with best practice for systematic reviews (Higgins et al., 2022).

The use of PICO allowed each study to be mapped consistently across:

- Population – stroke survivors at varying stages of recovery (acute, subacute, chronic), with sample sizes ranging from 8 to 214 participants.
- Intervention – a wide spectrum including gait training, orthotic prescription, functional electrical stimulation (FES), robotic-assisted therapy, treadmill training, and adjunctive technologies.

- Comparison – varied from usual care or no intervention, to pre–post designs without a control group, through to controlled or randomised comparisons.
- Outcomes – typically walking speed, step length, and cadence, but in some cases extended to propulsion, trailing-limb angle, muscle synergy analysis, or EMG-derived measures.

This structure ensured consistency in data handling and allowed for direct comparison of key study elements across a heterogeneous evidence base. It also facilitated transparent separation of studies reporting only functional outcomes from those incorporating mechanistic or biomechanical measures, a distinction central to the aims of this review.

The PRISMA diagram (Figure 1) summarises the flow of records through the review. Out of an initial 4,212 records identified through database searching, 3,443 remained after removal of duplicates. Addition of search parameters – date range 2004 to 2014, English language, adult cohorts and non-empirical publications excluded 3,130 papers, leaving 313 studies for full-text review. Full text papers were then screened using the stated inclusion and exclusion criteria. Of these, 238 were excluded for reasons such as inappropriate population ($n = 74$), intervention outside scope ($n = 92$), lack of gait-related outcomes ($n = 51$), or insufficient methodological detail ($n = 21$). This process yielded the final 75 studies included in the synthesis.

The full results table is provided in Appendix A. While this chapter presents the methodological overview and quantitative scope of the evidence base, critical appraisal of findings and their integration with mechanistic models of gait recovery are developed in later sections (see Section 2.7).

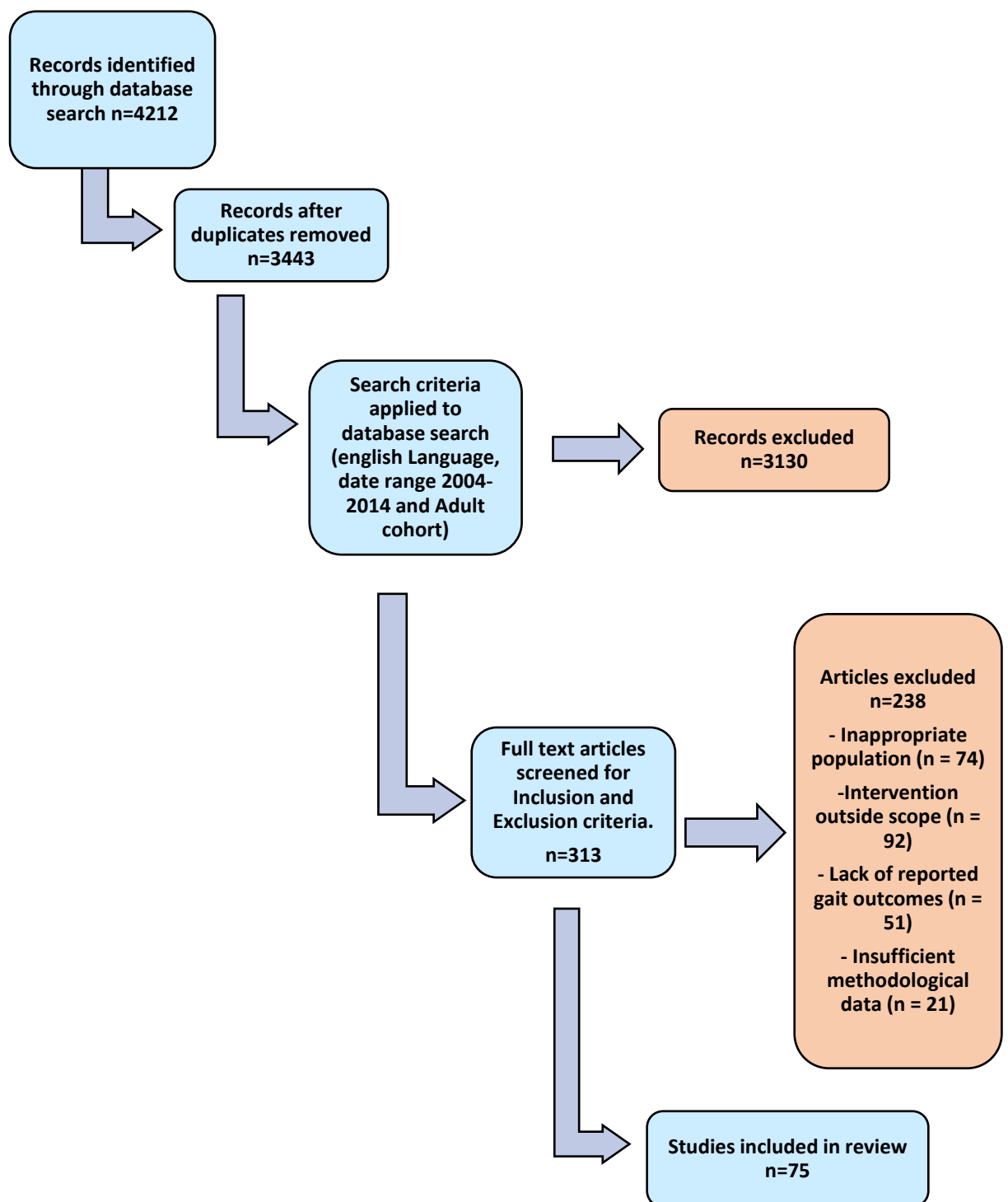


Figure 1 - PRISMA Flow diagram to illustrate studies included in Literature Review

2.6.4.1 Narrative Synthesis Approach

Given the heterogeneity of interventions, outcomes, and study designs, meta-analysis was not feasible. Instead, a structured narrative synthesis was employed, consistent with best practice for complex rehabilitation research (Popay et al., 2006). Extracted data were tabulated according to the PICO framework (Appendix A), which provided the basis for thematic grouping.

Studies were clustered into domains reflecting propulsion-targeted training, orthotic interventions, EMG and synergy analyses, and functional predictors of gait outcomes.

Within each theme, findings were analysed in relation to study design, population, intervention type, and outcome measures. Findings were then disseminated using a narrative approach which allowed integration of evidence across diverse methodologies, while maintaining transparency in distinguishing functional outcomes (e.g., walking speed) from mechanistic measures (e.g., propulsion, synergy structure, shank-to-vertical angle).

2.6.4.2 PRISMA-S Compliance

In addition to following PRISMA 2020 reporting standards for systematic reviews, the search strategy was documented in accordance with the PRISMA-S (Search extension) checklist (Rethlefsen et al., 2021). This ensured that full search strategies for each database, including Boolean operators, truncations, filters, and date limits, were archived and reported in Appendix A. Adherence to PRISMA-S therefore strengthens confidence in the comprehensiveness of the literature base underpinning this systematic review.

2.6.5 Full PICO Results Table

The full table of included papers (n=75) can be found in Appendix A. The table identifies individual Study Type, Populations, Key Measures of review and Key findings for each paper alongside the Authors and APA 7th reference.

2.6.6 Search Protocol Summary

This framework ensured that the review focused on post-stroke gait and its neuromuscular underpinnings, while excluding peripheral or tangential research that could dilute the findings.

This systematic process ensured that the final review represents the most relevant and methodologically robust literature in the field of post-stroke gait rehabilitation using EMG and gait analysis with an additional reflection on the use of Orthoses, specifically Ankle Foot Orthoses (AFOs) in the post-Stroke rehabilitation journey.

2.7 Literature Review Findings

The findings of the systematic literature review are organised thematically to reflect the main areas of evidence relevant to the thesis aims. Rather than presenting the included studies in chronological or methodological order, the discussion is structured around the key concepts that emerged from the review: functional gait impairment, neuromuscular control, clinically relevant gait indicators, rehabilitation practice, targeted intervention approaches, and physiotherapy-led management. This structure was chosen to allow the evidence to be synthesised in relation to the central concern of the thesis: how post-stroke gait can be assessed and interpreted in a way that captures both biomechanical performance and underlying neuromuscular function.

While acknowledging there is understandable repetition of papers between thematic sections, the following is an overview of the section structure: section 2.7.1 first considers functional gait post-stroke, drawing on 28 papers to establish the principal gait deviations, compensatory strategies, prognostic factors, and clinical implications identified across the evidence base. Section 2.7.2 then focuses on muscle activation and EMG evidence, discussing 17 papers concerned with synergy structure, muscle-specific abnormalities, co-contraction, and methodological issues in EMG interpretation.

Section 2.7.3 examines commonly reported gait indicators, including gait speed, shank-to-vertical angle, trailing-limb angle, and paretic propulsion, drawing on 27 papers. Section 2.7.4 reviews common practices in stroke rehabilitation, using 10 papers to consider current intervention approaches and their limitations. Section 2.7.5 then explores rehabilitation approaches specifically targeting gait and muscle activity, drawing on 13 papers. Finally, Section 2.7.6 considers physiotherapy practice post-stroke, using 9 papers to situate these findings within wider rehabilitation delivery.

This thematic structure enables the review to move from broad description of post-stroke gait impairment towards more specific consideration of mechanistic assessment and intervention. It also supports critical comparison across studies by highlighting where evidence is consistent, where methodological limitations remain, and where clinically important gaps persist.

2.7.1 Functional Gait post-Stroke

Post-stroke gait dysfunction is multidimensional, characterised by alterations in spatiotemporal symmetry, kinematic and kinetic deviations, abnormal muscle synergies, and secondary musculoskeletal adaptations. Across the included studies a consistent pattern emerged: survivors walk more slowly, with reduced step length and cadence (Patterson et al., 2008; Wonsetler & Bowden, 2017), but the mechanisms underpinning this “slowness” differed across cohorts and methodologies. Critically, gait impairment cannot be reduced to a single variable, such as speed, because spatial and temporal asymmetries, propulsion deficits, and neuromuscular coordination each contribute differently to function and recovery potential.

2.7.1.1 *Spatial and Temporal Asymmetries*

Stroke survivors demonstrate both spatial asymmetry (unequal step/stride lengths) and temporal asymmetry (unequal stance or swing times). Stanhope, Knarr, Reisman, and Higginson (2014) showed that temporal asymmetry - specifically reduced stance time on the paretic limb and prolonged swing, was more tightly associated with dynamic instability than spatial asymmetry. By contrast, Bowden et al. (2006) highlighted propulsion asymmetry as the dominant limitation, with paretic limbs in severely impaired walkers contributing only ~16% of total propulsion versus ~49% in milder hemiparesis. Together

these findings illustrate that while both spatial and temporal asymmetries are prevalent, their functional relevance differs temporal asymmetry reflects instability and reliance on the non-paretic limb, whereas spatial/propulsion asymmetries primarily constrain walking speed and endurance. Rehabilitation therefore requires domain-specific targeting rather than generic “symmetry restoration.”

2.7.1.2 Propulsion and Trailing-Limb Mechanics

The question of what drives propulsion deficits has been addressed in several of the reviewed studies. Peterson, Cheng, Kautz, and Neptune (2010) used forward-dynamics modelling to show that leg extension angle—closely linked to trailing-limb angle (TLA)—was the strongest predictor of propulsive impulse, while premature hip flexor activity reduced propulsion. This contrasts with clinical emphasis on ankle plantarflexor weakness, suggesting that deficits in coordination of limb extension may matter more than strength alone. Hsiao, Knarr, Higginson, and Binder-Macleod (2015a) and Hsiao, Knarr, Pohlig, Higginson, and Binder-Macleod (2015b) confirmed that training-related increases in propulsion were mediated primarily by TLA rather than ankle moment, and that speed improvements closely tracked increases in propulsion. Knarr, Kesar, Reisman, Binder-Macleod, and Higginson (2013) further demonstrated in simulations that improved plantarflexor timing induced forward centre-of-mass acceleration, consolidating the mechanistic link.

These convergent findings reinforce that propulsion asymmetry is not merely a by-product of weakness but reflects impaired inter-segmental coordination, with TLA emerging as a clinically useful, modifiable target.

2.7.1.3 Synergy Patterns and Neuromuscular Coordination

Electromyographic (EMG) evidence from the review provides a neural-control perspective. Clark, Ting, Zajac, Neptune, and Kautz (2010) showed that stroke survivors express fewer or merged muscle synergies compared with healthy adults, with greater merging associated with slower walking and reduced coordination complexity. Cheung et al. (2012) demonstrated that altered synergies map onto cortical damage, strengthening the argument that synergy analysis reflects neurophysiological impairment rather than simply functional limitation. Barroso et al. (2017) extended this by showing that combined synergy, kinematic, and kinetic features predicted walking impairment more accurately than the Fugl–Meyer scale.

Collectively, these studies argue that abnormal synergy structure is a mechanistic bottleneck in post-stroke gait and a potentially more sensitive predictor of recovery than traditional impairment scores. However, methodological heterogeneity, including how synergies are extracted, the number of muscles included, and whether treadmill or overground walking is assessed, limits comparability across studies.

2.7.1.4 Gait deviations Post-Stroke

Clinically, foot drop refers to impaired dorsiflexion during swing due to tibialis anterior weakness or poor activation, producing compensatory high-stepping gait and risk of tripping. Toe drag, defined as insufficient clearance of the toes during swing, is often its direct consequence. Burpee and Lewek (2015) demonstrated how marginal the difference can be: the angular distinction between a “trip” and a “clear” step was only $\sim 10^\circ$, implying that small variations in dorsiflexion or limb trajectory can cause falls. These findings highlight both the vulnerability of stroke survivors to clearance errors and the risk that orthotic devices or fixed-timing FES, if improperly tuned, could overshoot and suppress the very propulsion required for efficient gait.

Stiff-knee gait (SKG) is another common deviation. Yuan, Ge, Du, and Xia (2019) showed that rectus femoris–biceps femoris co-contraction ratios were significantly higher on the paretic side (0.71 vs 0.45, $p < .01$), and that this was associated with reduced knee flexion and worse impairment. Yet Campanini, Merlo, and Damiano (2013) reported that ~85% of SKG cases were driven primarily by inadequate push-off rather than rectus femoris overactivity, highlighting the need for patient-specific assessment. This explains why interventions targeting rectus femoris with botulinum toxin are variably effective and why dynamic EMG should guide injection decisions (Merlo & Campanini, 2019).

2.7.1.5 Developing Deformity

Equinovarus deformity is characterised by persistent ankle plantarflexion and inversion (supination), compromising heel strike and stance stability. This deformity amplifies both spatial and temporal asymmetries by forcing weight-bearing onto the lateral border of the foot. Mulroy, Eberly, Gronley, Weiss, and Newsam (2010) demonstrated that different AFO designs alter ankle/knee kinematics and EMG activity, while Bregman, Harlaar, Meskers, and de Groot (2012) showed that spring-like AFOs reduced walking energy cost by ~10% by substituting for ankle positive work. Harper et al (2014) extended this line of evidence, showing that tuning AFO stiffness modifies both energy cost and kinematic alignment. Collectively, these studies confirm that management of equinovarus requires careful orthotic tuning (see 2.4 for a detailed orthotic analysis).

2.7.1.6 Spasticity and Tone

Spasticity is defined as velocity-dependent hyperexcitability of the stretch reflex, while muscle tone refers to resistance to passive stretch in a relaxed

state (Sheean & McGuire, 2009; Dietz & Sinkjaer, 2007). In post-stroke gait, increased extensor tone and spastic co-contraction restrict knee flexion and ankle dorsiflexion, thereby aggravating clearance problems and SKG. Evidence from included studies shows that inappropriate rectus femoris activity can limit swing (Yuan et al., 2019), but classification studies caution that reduced propulsion often plays a larger role (Campanini et al., 2013). Thus, spasticity is only part of the picture: coordination deficits and propulsion failure must be addressed in parallel.

2.7.1.7 Prognostic Significance and Clinical Implications

Early clinical predictors consistently emphasise the central role of motor control. In the EPOS study, Veerbeek, Van Wegen, Harmeling-Van der Wel, and Kwakkel (2011) reported that sitting balance combined with any voluntary leg movement within 72 hours predicted independent walking at 6 months with ~98% accuracy. Stinear, Lang, Zeiler, and Byblow (2017) confirmed that trunk control and lower-limb strength are core determinants of recovery. These findings dovetail with mechanistic data: propulsion and synergy deficits (Bowden et al., 2006; Clark et al., 2010) may explain why early motor signs are such strong predictors of outcome. Yet limitations remain: many studies exclude AFO users (Burpee & Lewek, 2015; Jonkers, Delp, & Patten, 2009), potentially biasing results toward higher-functioning groups.

Clinically, these insights imply that effective rehabilitation must go beyond gait speed. Gains in laboratory metrics may not generalise to real-world function: Mahendran, Kuys, and Brauer (2016) found that indoor walking improvements did not extend to outdoor activity. The implication is that interventions should target propulsion and clearance mechanics while simultaneously measuring ecological outcomes such as community walking (see 2.3.3 for gait speed/TLA/PP indicators; 2.3.5 for targeted propulsion-training evidence).

2.7.2 Muscle Activation Patterns in Post-Stroke Gait (EMG Evidence)

Electromyography (EMG) provides an indispensable window into the neuromuscular mechanisms underpinning post-stroke gait. Unlike purely kinematic or kinetic measures, EMG captures the timing, amplitude, and coordination of muscle activation, exposing both selective weakness and systemic disruption of motor control. The included studies consistently demonstrate that survivors do not simply “scale down” normal activity patterns but instead show qualitatively altered coordination, abnormal co-contractions, and reduced modularity. These findings have direct implications for prognosis and therapy, though methodological heterogeneity complicates synthesis.

2.7.2.1 Synergy Structure and Motor Control

A central theme in post-stroke motor control research is that the complexity of muscle coordination, the modular “synergy structure”, reflects the integrity of neural control networks. Foundational work by Clark, Ting, Zajac, Neptune, and Kautz (2010) demonstrated that individuals post-stroke typically express *fewer or merged muscle synergies* compared with healthy adults. This reduction in modularity correlated strongly with slower gait speed ($r \approx .84$) and poorer inter-limb coordination, implying that synergy decomposition provides a compact representation of impaired neural command flexibility. Cheung et al. (2012) subsequently confirmed that altered synergy structure aligns with lesion location and extent on neuroimaging, strengthening the interpretation of muscle synergies as biomarkers of central rather than purely peripheral dysfunction.

However, more recent evidence challenges the simplicity of this relationship. Recent evidence challenges the assumption that muscle synergy number is a stable marker of post-stroke motor impairment. Santuz et al. (2017) demonstrated that extracted synergy number varied significantly according to

factorisation approach and filtering strategy, rather than reflecting physiology alone. Across locomotor conditions, inverse-Gaussian NMF required approximately one fewer synergy than Gaussian NMF to reconstruct the same EMG signals, with differences significant across level walking, incline walking, shod running, and barefoot running ($p < .001$). For example, level walking required 4 ± 1.0 synergies using Gaussian NMF compared with 3 ± 1.1 using inverse-Gaussian NMF, while incline walking required 5 ± 0.8 versus 4 ± 0.9 synergies, respectively. This directly undermines simplistic interpretations of synergy count, as the same locomotor data can produce different modular structures depending on processing decisions.

Routson et al. (2013) provide a useful contrast because their findings support the functional relevance of muscle modules but also show that module quality is more informative than module number alone. In participants with four modules before and after therapy, self-selected speed improved significantly ($p = .0114$), pre-swing leg angle also improved ($p = .0440$), and plantarflexor module timing improved ($p = .0132$). However, changes in propulsion asymmetry only approached significance ($p = .1121$), indicating that improved modular timing did not translate uniformly into all biomechanical outcomes. This suggests that rehabilitation-related change is more complex than simply restoring a greater number of synergies.

Routson et al. also showed that some impairments persisted despite therapy. Plantarflexor timing was significantly impaired pre-therapy compared with controls ($p = .0004$), but after therapy was no longer significantly different ($p = .65$). In contrast, tibialis anterior/rectus femoris module timing remained significantly different from controls after therapy ($p < .0001$), and plantarflexor module composition also remained impaired ($p = .0160$). This pattern is important because it shows selective improvement rather than global normalisation: some aspects of neuromuscular organisation responded to rehabilitation, while others remained abnormal.

Taken together, Santuz et al. (2017) and Routson et al. (2013) expose a central weakness in the synergy literature: synergy number is often treated as a proxy for neural recovery, yet it is sensitive to methodological choices and insufficient to explain functional change on its own. A more defensible interpretation is that post-stroke recovery should be evaluated through the timing, composition, and task-specific adaptability of muscle modules, rather than through module count alone. This supports a more cautious reading of earlier work linking reduced modularity to impaired gait and reinforces the need for transparent EMG processing pipelines in future studies.

Comparatively, Barroso et al. (2017) integrated synergy metrics with kinematic and kinetic variables, demonstrating improved prediction of gait impairment compared with the Fugl–Meyer scale alone ($R^2 = 0.72$ vs 0.55). More recent work has challenged the tendency to interpret reduced synergy number as a direct marker of impairment severity. Santuz et al. (2017) demonstrated that extracted synergy number varied significantly depending on factorisation strategy and preprocessing choices, with inverse-Gaussian non-negative matrix factorisation consistently identifying fewer synergies than Gaussian approaches across locomotor tasks ($p < .001$). These findings suggest that apparent reductions in modular complexity may partly reflect methodological artefact rather than true neurophysiological deterioration.

Similarly, Routson et al. (2013) showed that rehabilitation-related improvements were more strongly associated with changes in module timing and organisation than with module number itself. Following therapy, self-selected walking speed improved significantly ($p = .0114$) alongside improvements in plantarflexor timing ($p = .0132$), despite persistent abnormalities in other module characteristics. Collectively, these studies suggest that synergy quality, timing, and adaptability may be more clinically meaningful than absolute module count alone.

From a critical standpoint, this newer body of literature represents an important methodological advance because it moves beyond descriptive comparisons of “more versus fewer” synergies and interrogates how analytical decisions influence findings. Santuz et al. (2017) explicitly demonstrated that preprocessing pipelines and factorisation thresholds materially alter extracted modular structures, exposing a major source of inter-study inconsistency often overlooked in earlier literature. Routson et al. (2013), meanwhile, linked module-specific changes to functional outcomes, strengthening the clinical relevance of synergy analysis.

However, important limitations remain. Most studies continue to rely on surface EMG, restricting analysis to superficial musculature while excluding deeper stabilising muscles and intersegmental coordination patterns. Sample sizes also remain modest, limiting statistical power and external validity. In addition, many synergy studies remain cross-sectional or short-term intervention designs, preventing causal inference regarding whether altered modularity reflects neural damage, compensatory adaptation, or recovery-related reorganisation.

The evolving evidence base therefore indicates that post-stroke motor control deficits arise not simply from reduced modular complexity, but from disrupted timing, altered organisation, and reduced adaptability within locomotor modules. Rather than treating synergy number as a static biomarker of lesion severity, current evidence (Clark et al., 2010; Cheung et al., 2012; Routson et al., 2013; Santuz et al., 2017) supports a more dynamic interpretation in which recovery depends on restoring flexible and context-sensitive coordination patterns. Methodologically, the literature now recognises that variability in reported synergy structure frequently reflects differences in EMG preprocessing, normalisation procedures, and gait conditions as much as genuine neurophysiological disparity.

Consequently, synergy analysis should be viewed as a powerful but highly parameter-sensitive tool whose interpretability depends on methodological transparency, reproducible processing pipelines, and integration with complementary biomechanical measures.

2.7.2.2 Muscle-Specific Abnormalities

At the level of individual muscles, the most robust abnormality across the literature is reduced tibialis anterior activity during swing, resulting in foot drop and toe drag (see 2.3.1 for clinical definitions and orthotic implications). Abnormalities also extend to the quadriceps–hamstring complex. Yuan, Ge, Du, and Xia (2019) found significantly higher rectus femoris–biceps femoris co-contraction ratios on the paretic side (0.71 vs 0.45; $p < .01$), which were associated with reduced knee flexion and worse functional impairment. This aligns with clinical presentations of stiff-knee gait (SKG).

However, Campanini, Merlo, and Damiano (2013) cautioned that rectus femoris over-activity is not the primary driver in most cases: ~85% of SKG presentations were attributable to inadequate push-off rather than excessive rectus femoris braking. This distinction is clinically important, as it helps to explain why botulinum toxin injections to the rectus femoris yield variable outcomes. Merlo and Campanini (2019) reinforced this point, showing that dynamic EMG could prevent unnecessary injections in up to 81% of cases, highlighting the clinical utility of EMG in tailoring treatment.

2.7.2.3 Speed Modulation and Coordination limits

Speed manipulation experiments illustrate the persistence of coordination constraints. Buurke, Lamothe, van der Woude, and den Otter (2016) reported

that when stroke survivors attempted to increase speed, abnormal co-activation patterns persisted, limiting functional gains. These findings resonate with propulsion-focused studies showing that speed improvements only occur when trailing-limb angle improves (Hsiao, Knarr, Higginson, & Binder-Macleod, 2015a; Hsiao, Knarr, Pohlig, Higginson, & Binder-Macleod, 2015b). Thus, abnormal EMG patterns are not passive epiphenomena but active bottlenecks that constrain the capacity to exploit speed training.

2.7.2.4 Comparative Studies of Muscle Activity in Stroke and Healthy Populations

Comparative analyses of stroke survivors and healthy adults provide an essential lens for understanding the scale of neuromuscular disruption in gait and for differentiating between genuine recovery and compensatory strategies. Surface EMG has been the principal tool in these comparisons, enabling evaluation of both muscle-specific activity and higher-order modular control.

At the muscle level, deviations from healthy gait are consistent. Tibialis anterior activation during swing is typically reduced in stroke survivors, resulting in inadequate dorsiflexion and recurrent foot drop, while plantarflexor activation in late stance is delayed or absent, undermining forward propulsion (Boudarham et al., 2013; Bowden, Balasubramanian, Neptune, & Kautz, 2006). Abnormal co-contraction between quadriceps and hamstrings further stiffens the limb, increasing energy cost and reducing efficiency (Clark, Ting, Zajac, Neptune, & Kautz, 2010). These deficits link directly to functional limitations described earlier in Sections 2.7.1 (spatial and temporal asymmetry) and 2.7.3.1 (speed outcomes).

Synergy-level analyses extend this understanding. Healthy adults typically organise walking into four to five distinct synergies, reflecting modular neuromuscular control. Stroke survivors often demonstrate fewer or merged

synergies, indicating reduced adaptability (Clark et al., 2010; Cheung et al., 2012). Importantly, Cheung et al. (2012) connected synergy merging to cortical damage, while Barroso et al. (2017) showed that integrating synergy data with kinematic and kinetic measures predicted impairment more accurately than traditional clinical scales. This underscores the value of synergy analysis as both a biomarker of impairment and a complement to biomechanical metrics such as trailing limb angle and paretic propulsion (see Sections 2.7.3.3–2.7.3.4).

Specific gait deviations illustrate how muscle- and synergy-level findings intersect. Stiff-knee gait (SKG) is one such example. Elevated rectus femoris–biceps femoris co-contraction ratios (Yuan, Ge, Du, & Xia, 2019) and inappropriate vastus intermedius activation (Mazzoli et al., 2018) have been linked to limited knee flexion in swing. Yet classification analyses suggest that in many cases, SKG reflects insufficient push-off rather than excessive rectus femoris braking (Campanini, Merlo, & Damiano, 2013). This interpretation aligns with the propulsion-focused mechanisms already discussed in Section 2.7.3.4, reinforcing the point that visible gait deviations often arise from upstream deficits in propulsion rather than isolated joint dysfunction.

Taken together, comparative studies reveal that stroke gait cannot be understood as a slowed version of normal walking. Instead, it is characterised by disrupted neuromuscular coordination, reduced synergy complexity, and compensatory co-contraction patterns. Observed deviations such as foot drop or SKG therefore need to be interpreted as outcomes of broader propulsion and coordination deficits, which are best evaluated with integrated metrics of biomechanics and modular control.

This places comparative EMG evidence in direct dialogue with earlier discussions of gait speed, propulsion, and trailing limb angle, and highlights

the importance of combining these perspectives for a complete picture of recovery.

2.7.2.5 Methodological Caveats in EMG Interpretation

Several included studies highlight the challenges of interpreting EMG amplitude and timing across trials. Baudet et al. (2014) demonstrated that crosstalk and processing choices (e.g., filtering, normalisation) significantly influence reported activation levels, while Den Otter, Geurts, Mulder, and Duysens (2006) showed that patients may achieve functional recovery without normalisation of temporal activation patterns. Neptune, Clark, and Kautz (2009) argued that activation timing is generally more reliable than amplitude for cross-study comparison, a point borne out by the variability of amplitude findings across the corpus. These methodological insights caution against equating “normalised EMG” with “recovery” and highlight the need for transparent pipelines.

2.7.2.6 Clinical and Research Implications

Taken together, EMG evidence from the literature review demonstrates that post-stroke gait deficits are rooted in disrupted neuromuscular coordination rather than weakness alone. Clinically, this explains why interventions focusing solely on strength (e.g., isolated dorsiflexor exercises) often fail to restore efficient gait. Instead, therapies should target coordination, propulsion, and synergy restructuring (see 2.7.5 for evidence on FastFES and task-specific propulsion training). At the same time, methodological heterogeneity limits meta-analysis and generalisability.

In summary, EMG analysis adds a critical dimension to our understanding of post-stroke gait, exposing coordination deficits invisible to kinematics alone.

Yet these insights are only as robust as the methods underpinning them. The clinical challenge is to translate mechanistic findings, such as synergy loss and abnormal co-activation, into personalised rehabilitation strategies that address both the neural control deficits and the biomechanical consequences identified elsewhere in this review.

2.7.3 Common Indicators of Gait Function

A wide range of outcome measures are used to characterise gait function post-stroke. While gait speed remains the most common, evidence from the included studies demonstrates that it is often an incomplete surrogate, failing to capture underlying biomechanical deficits or compensatory mechanisms. Alternative indicators such as shank-to-vertical angle (SVA), trailing-limb angle (TLA), and paretic propulsion (PP) have emerged as more mechanistically sensitive measures. Prognostic studies additionally highlight early motor signs and endurance tests as robust predictors of long-term recovery. This section synthesises evidence across these domains.

2.7.3.1 *Gait Speed*

Gait speed is one of the most widely used clinical and research indicators of post-stroke walking ability. Across the included studies, gait speed emerged as both a strength and a limitation when employed as a rehabilitation outcome. Its strengths lie in simplicity, reproducibility, and sensitivity to change. For example, Bowden et al. (2013) used gait speed to stratify responders versus non-responders to locomotor training, demonstrating that speed captured meaningful between-group differences in adaptation. Likewise, Kerr et al. (2020) showed that improvements in walking speed over 6 months reflected genuine progress in functional mobility, even if thresholds for community ambulation were not fully reached.

However, several studies demonstrate the limitations of gait speed when interpreted in isolation. Mahendran, Kuys, and Brauer (2016) found that gains in indoor walking speed failed to translate into greater outdoor participation, illustrating that speed lacks ecological validity as a proxy for real-world reintegration. Burpee and Lewek (2015) added nuance by showing that externally prescribed speed altered gait patterns differently than self-selected speed, underscoring that speed changes can be confounded by task demands rather than representing recovery per se. Similarly, Kerrigan, Karvosky, and Riley (2001) noted that altering walking speed in healthy participants induced atypical gait adaptations, implying that speed effects in stroke must be contextualised against biomechanical constraints.

Another limitation is that gait speed cannot identify the mechanistic drivers of change. Peterson, Cheng, Kautz, and Neptune (2010) showed that leg extension angle and hip flexor impulse explained propulsive deficits more directly than speed, while Hsiao, Knarr, Higginson, and Binder-Macleod (2015a) demonstrated that increases in propulsion were mediated by trailing-limb angle rather than ankle moment. Thus, two individuals with similar speed gains may rely on very different biomechanical strategies—compensation in one case, recovery in another.

Critically, studies combining gait speed with endurance or propulsion outcomes demonstrate greater explanatory power. Selves, Stoquart, and Lejeune (2020) found the 6-minute walk test (6MWT) better discriminated rehabilitation outcomes than speed alone, while Fulk, He, Boyne, and Dunning (2017) showed that endurance thresholds provided superior prediction of community ambulation. Awad, Reisman, Pohlig, and Binder-Macleod (2016) reported that increases in speed after FastFES were paralleled by increases in paretic propulsion, illustrating that when combined, speed and

biomechanical measures can clarify whether improvements represent true recovery of function.

In stroke survivors, self-selected walking speed (SSWS) is consistently reduced compared with healthy adults, and this reduction stratifies functional ambulation. Historically Perry, Garrett, Gronley, and Mulroy (1995) established cut-offs of <0.40 m/s (household ambulation), 0.40–0.80 m/s (limited community), and >0.80 m/s (community), thresholds that remain widely applied and are detailed here to enable comparison of data.

Evidence from recent cohorts reinforces both the clinical utility and the limitations of SSWS. An, Lee, Shin, and Lee (2015) demonstrated that a threshold of 0.87 m/s predicted community ambulation with high accuracy (AUC 0.87), while Fulk, He, Boyne, and Dunning (2017) showed that the 6-minute walk test (6MWT) outperformed SSWS in discriminating between community and household walkers. Selves, Stoquart, and Lejeune (2020) confirmed that endurance-based outcomes such as the 6MWT provide superior predictive validity for community walking compared to SSWS alone.

Interventional studies further highlight variability in speed response. Bowden, Behrman, Neptune, Gregory, and Kautz (2013) reported that responders to locomotor training improved SSWS by $+0.27 \pm 0.12$ m/s and 6MWT distance by $+196 \pm 200$ m, whereas non-responders showed minimal change, emphasising that speed gains do not occur uniformly. Kerr et al. (2020) found that although mean gait speed increased significantly at 6 months, values remained below the 0.80 m/s community threshold, suggesting persistent functional restrictions despite improvement. Mahendran, Kuys, and Brauer (2016) added that indoor speed gains did not translate into increased outdoor/community activity, underscoring ecological limits.

Methodological differences across studies also complicate interpretation. Burpee and Lewek (2015) tested therapist-selected walking speeds, while others (Kerr et al., 2020; Mahendran et al., 2016; Selves et al., 2020) measured self-selected speeds, altering ecological validity.

Critically, increases in SSWS cannot be assumed to represent qualitative recovery. While SSWS is sensitive to change and easy to measure, it fails to reveal whether patients rely on compensations such as increased cadence or asymmetrical propulsion. The convergent evidence from endurance-based outcomes (Fulk et al., 2017; Selves et al., 2020) and propulsion-focused measures (see 2.7.3.4) suggests that speed must be interpreted in a multidimensional framework.

Taken together, the included studies demonstrate that gait speed is a valid, reproducible, and sensitive marker of functional improvement, but it is insufficient when interpreted in isolation. While SSWS provides a convenient benchmark for classifying ambulation status and tracking change, it cannot capture the biomechanical strategies underlying improvement or the ecological transfer of gains into community participation.

Evidence from Mahendran, Kuys, and Brauer (2016) and Kerr et al. (2020) illustrates that increases in speed may not translate into outdoor activity or community walking, while Burpee and Lewek (2015) caution that speed changes can reflect altered task demands rather than genuine recovery. Mechanistic analyses reinforce this limitation: Peterson, Cheng, Kautz, and Neptune (2010) and Hsiao, Knarr, Higginson, and Binder-Macleod (2015a) showed that propulsion mechanics, not speed per se, explain functional deficits and improvements.

Accordingly, gait speed should be triangulated with endurance measures such as the 6MWT (Fulk, He, Boyne, & Dunning, 2017; Selves, Stoquart, & Lejeune,

2020) and biomechanical outcomes including paretic propulsion (Bowden, Balasubramanian, Neptune, & Kautz, 2006; Awad, Reisman, Pohlig, & Binder-Macleod, 2016) and trailing-limb angle (Hsiao, Knarr, Higginson, & Binder-Macleod, 2015a). This multidimensional approach allows researchers and clinicians to distinguish compensatory strategies from true neuromechanical recovery and to better predict meaningful reintegration into community walking.

2.7.3.2 Shank to Vertical Angle (SVA)

The shank-to-vertical angle (SVA) describes the inclination of the tibia relative to vertical during stance and has emerged as a clinically modifiable determinant of gait mechanics. SVA influences the line of action of the ground-reaction force (GRF) relative to the knee and hip, thereby shaping extension moments, rocker progression, and ultimately the efficiency of stance. In post-stroke rehabilitation, it is most frequently addressed through the alignment and tuning of ankle–foot orthosis/footwear combinations (AFO-FCs), which provide an external means of adjusting shank progression.

Empirical evidence from the included full-text studies demonstrates that changes in SVA, whether deliberate or incidental, have measurable effects on both kinematics and function. In a within-subject study of thirty post-stroke adults, Mulroy, Eberly, Gronley, Weiss, and Newsam (2010) compared four different AFO designs against a shoes-only condition. Across all AFOs, swing- and early-stance dorsiflexion was improved, but designs that permitted ankle motion had distinct additional effects: they altered knee kinematics and changed muscle activity patterns in the soleus and tibialis anterior. These findings highlight that the benefits of AFOs are not uniform but depend on how the device interacts with shank alignment and muscle recruitment.

The influence of SVA on energetic efficiency is evident in the work of Bregman, Harlaar, Meskers, and de Groot (2012), who demonstrated that a spring-like AFO reduced the net metabolic cost of walking by approximately 10%. They argued that this occurred because the device assumed part of the ankle's positive work and facilitated smoother shank progression. In a similar vein, Harper et al (2014) systematically manipulated AFO stiffness and found consequential changes in both ankle–knee kinematics and overall walking economy. These findings reinforce the notion that SVA is not an abstract angle but a clinically actionable lever: when tuned appropriately, it can lower effort and improve mechanical efficiency.

Beyond energy and joint angles, Bleyenheuft, Deltombe, and Detrembleur (2013) illustrated how SVA-related tuning may support inter-segmental coordination. Their study showed that a dynamic AFO led to a more natural and efficient movement pattern in the paretic limb - a feature not achieved by a prefabricated rigid AFO. This implies that achieving a physiologically appropriate SVA facilitates not only joint-level mechanics but also the integration of limb segments into more coordinated movement patterns.

Functional correlates were provided by Kerr et al (2020), who observed that tibia-to-vertical angle during stance was significantly correlated with walking speed and functional walking capacity in community-dwelling stroke survivors. This suggests that SVA can act as a bridge between biomechanical control and clinically meaningful outcomes. However, the study did not include concurrent orthotic intervention, leaving open the question of how orthotic tuning of SVA might further enhance functional gains (see Section 2.4).

Cross-analysis of these studies points to a coherent message: SVA is a modifiable parameter that influences gait mechanics at multiple levels: joint kinematics, muscle activity, energy cost, coordination, and functional capacity. Yet, as seen in section 2.7.2, methodological heterogeneity limits

generalisation. Device designs differ (rigid, dynamic, spring-like), outcome measures vary (energy cost, EMG, coordination), and most propulsion-oriented training studies excluded participants using AFOs (e.g., Burpee & Lewek, 2015; Jonkers, Delp, & Patten, 2009). This exclusion creates a translational gap between research protocols and real-world practice, where AFO prescription is common. The implication is that while we know SVA tuning is effective in controlled contexts, its interaction with active gait retraining remains underexplored.

From a clinical perspective, SVA should be viewed as complementary to other gait indicators. For example, improvements in SVA during stance should, in theory, align with gains in trailing-limb angle (TLA; see Section 2.7.3.3) and paretic propulsion (Section 2.7.3.4), as all three variables converge on the restoration of efficient forward progression. Equally, changes in SVA have implications for orthotic practice, where device alignment and stiffness directly dictate shank inclination. Integrating SVA with endurance and propulsion measures offers a more comprehensive assessment of recovery, helping clinicians distinguish between compensatory stability and genuine restoration of efficient mechanics.

In summary therefore, the available evidence establishes SVA as a mechanistically important and clinically adjustable determinant of gait performance. Well-tuned SVA has been shown to reduce energy cost, enhance knee control, and improve coordination, but the lack of studies combining SVA tuning with modern propulsion-targeted rehabilitation remains a critical limitation. Addressing this gap could determine whether orthotic optimisation should be considered a stand-alone compensatory strategy or an active contributor to long-term recovery.

2.7.3.3 Trailing Limb Angle

The trailing limb angle (TLA), defined as the angle between a vertical reference and the line from the greater trochanter to the fifth metatarsal at terminal stance, has emerged as a pivotal kinematic determinant of propulsion in post-stroke gait. By dictating the orientation of the lower limb relative to the ground reaction force (GRF), TLA determines whether the GRF vector generates an anterior component that facilitates forward progression. In this way, TLA functions not merely as a descriptive parameter but as a mechanistic link between limb alignment and whole-body propulsion, complementing kinetic outcomes such as anterior–posterior propulsive impulse and paretic propulsion (see Section 2.7.3.4).

A modelling study by Peterson, Cheng, Kautz, and Neptune (2010) provided foundational evidence for the role of TLA in propulsion. Using musculoskeletal simulations based on experimental gait data from stroke survivors, they demonstrated that leg extension angle, a measure closely related to TLA, was the primary positive predictor of paretic propulsive impulse. Importantly, they also found that hip-flexor moment impulse was negatively associated with propulsion. This suggested that premature or excessive hip-flexor activity effectively “pulled” the limb off the ground too early, reducing the time and alignment necessary for effective push-off. These insights highlight a critical trade-off: without sufficient limb extension, the capacity of the plantarflexors to contribute to forward progression is constrained, even if muscle strength is adequate.

Clinical evidence aligns with these modelling insights. In a cohort study, Tyrell et al. (2011) measured TLA directly in post-stroke individuals and confirmed a strong positive correlation between larger TLA values and greater paretic propulsion. Their findings support the hypothesis that TLA is not just a biomechanical descriptor but a functionally relevant determinant of gait efficiency. By quantifying the direct relationship between limb extension and

propulsive force, Tyrell et al. (2011) offered empirical validation that TLA can serve as a clinical marker of recovery potential.

Experimental manipulations of walking speed further strengthen the evidence. Hsiao, Knarr, Higginson, and Binder-Macleod (2015a) examined how stroke survivors increased propulsion when walking faster. They found that the gains were driven predominantly by increases in TLA rather than by changes in ankle moment. This finding challenges the common clinical assumption that improving ankle plantarflexor strength alone is sufficient for propulsion recovery. Instead, it underscores the need for interventions that address both force generation and limb positioning, since even strong plantarflexors cannot produce forward propulsion if the limb is not oriented effectively.

In a longitudinal intervention trial, Hsiao, Knarr, Pohlig, Higginson, and Binder-Macleod (2015b) demonstrated that TLA is not only correlated with propulsion but is also a modifiable target. Over a 12-week gait training programme, participants showed significant improvements in TLA, which were directly associated with increases in paretic propulsion and walking speed. Crucially, these improvements were retained at follow-up, suggesting durable neuromechanical adaptations. The study highlights that changes in TLA can mediate functional gains, bridging the gap between impairment-level biomechanics and real-world walking outcomes.

Finally, Knarr, Kesar, Reisman, Binder-Macleod, and Higginson (2013) used forward-dynamics simulations to explore the causal mechanisms underpinning propulsion improvements. Their analysis showed that enhancing plantarflexor timing increased forward acceleration of the centre of mass (COM), but that this effect was contingent on improved leg extension. In other words, muscle activation patterns only translated into effective propulsion when the kinematic precondition of an adequate TLA was met. This simulation-based evidence

provides mechanistic clarity: TLA is the conduit through which muscular force is transformed into whole-body progression.

Cross-analysis of these studies reveals a consistent mechanistic pathway: adequate trailing-limb extension creates favourable alignment for anterior GRF, which in turn allows plantarflexor forces to accelerate the body forward. Conversely, premature hip-flexor activation or restricted TLA limits propulsion regardless of muscular capacity. The convergence of modelling, cross-sectional, experimental, longitudinal, and simulation evidence provides unusually robust triangulation for a biomechanical variable.

2.7.3.4 Paretic Propulsion (PP)

Paretic propulsion (PP), typically quantified as ratio of the propulsive impulse produced by the paretic limb to the combined anterior–posterior propulsive impulses of both limbs, providing a normalised measure of the paretic leg's contribution to forward propulsion (Roelker, Bowden, & Kautz, 2019) has gained traction as an outcome measure that captures a fundamental limitation in post-stroke gait. Whereas gait speed provides a global performance measure, PP isolates the kinetic contribution of the impaired limb, offering greater mechanistic specificity.

The seminal work of Bowden, Balasubramanian, Neptune, and Kautz (2006) established the clinical significance of PP. In their analysis of stroke survivors, they reported marked reductions in paretic-limb contribution to propulsion: individuals with severe hemiparesis generated only ~16% of total propulsion from the paretic limb, compared to ~49% in those with mild impairment. Furthermore, PP showed strong correlations with both walking speed ($r = .55$, $p < .001$) and clinical severity ($\rho = .74$, $p < .001$). These findings demonstrated not only that PP declines with impairment severity, but also that it mediates

functional capacity. The strength of these associations positioned PP as a potential surrogate marker for walking recovery.

Intervention studies support the modifiability of PP. In a randomised training study, Awad, Reisman, Pohlig, and Binder-Macleod (2016) examined propulsion-targeted locomotor training and showed that improvements in PP were accompanied by significant gains in self-selected walking speed and endurance. Specifically, participants classified as responders improved their speed by $+0.27 \pm 0.12$ m/s (well above the commonly accepted Minimal Clinically Important Difference (MCID) for post-stroke walking speed (~ 0.10 m/s; Tilson et al., 2010) and their 6MWT distance by $+196 \pm 200$ m, while non-responders exhibited minimal change. Crucially, improvements in PP explained a substantial portion of the variance in functional gains, underscoring its role as both a mechanistic driver and a predictor of rehabilitation outcomes.

Further evidence of PP's clinical utility comes from multivariate modelling. Barroso et al. (2017) combined kinetic measures such as PP with kinematic and synergy features to predict walking impairment, demonstrating that PP contributed significantly to the predictive accuracy of their models. Compared with conventional clinical scales, this integrated approach provided a more nuanced account of functional limitations. Such findings reinforce the view that PP cannot be considered in isolation but should be situated within a multidimensional framework that also includes measures like TLA (Section 2.7.3.3) and synergy structure (Section 2.7.2).

Mechanistic clarity was added by Hsiao, Knarr, Higginson, and Binder-Macleod (2015a), who showed that increases in propulsion during fast walking were predominantly driven by greater TLA rather than ankle moment. This finding implies that PP is not solely a reflection of muscle strength, but of the interaction between force capacity and limb alignment. Their subsequent

longitudinal work (Hsiao, Knarr, Pohlig, Higginson, & Binder-Macleod, 2015b) confirmed that PP improved in parallel with walking speed following training, again mediated by changes in TLA. Together, these studies underscore that PP is both a sensitive marker of impairment and a modifiable rehabilitation target, but that its underlying drivers must be addressed through coordinated kinematic and kinetic strategies.

The literature review highlights several converging themes. First, PP is strongly related to functional walking performance and scales with impairment severity, making it a robust mechanistic correlate of disability. Second, targeted training interventions can increase PP, and such increases are closely linked to functional gains in speed and endurance. Third, PP improvements often depend on concurrent increases in TLA, suggesting that kinetic measures must be interpreted alongside kinematic context. Finally, predictive modelling demonstrates that PP enhances explanatory power beyond clinical scales, making it a valuable inclusion in multidimensional assessment batteries.

Limitations in the literature include small sample sizes in most experimental and interventional studies, frequent exclusion of orthotic users (see Section 2.4), and heterogeneous methods for quantifying propulsive impulse. For example, some studies normalise PP to body weight while others use absolute force–time integrals, complicating cross-study comparisons. Additionally, while PP has been validated against laboratory outcomes (e.g., 6MWT, gait speed), fewer studies have examined its direct relationship with real-world activity or patient-reported outcomes, leaving its ecological validity partly untested.

Therefore, paretic propulsion provides a mechanistically rich complement to gait speed, capturing the impaired limb's kinetic contribution to forward progression. Across cross-sectional, intervention, and modelling studies, PP

emerges as both a marker of impairment severity and a modifiable rehabilitation target. However, its clinical utility depends on pairing it with kinematic indicators such as TLA and alignment parameters like SVA, and on extending its validation to community-based and patient-centred outcomes.

2.7.3.5 Predictors of Independent or Spontaneous Gait Recovery

Predicting which stroke survivors will regain independent gait has been a longstanding focus of neurorehabilitation, both to inform prognosis and to optimise allocation of rehabilitation resources.

Early bedside predictors remain the strongest clinical markers. In the large EPOS cohort (>150 participants), Veerbeek, Van Wegen, Harmeling-Van der Wel, and Kwakkel (2011) demonstrated that the combination of independent sitting balance and any voluntary leg movement within the first 72 hours predicted independent walking at six months with ~98% accuracy. This finding underscores the predictive weight of trunk control and early voluntary activation, which likely reflect preserved descending motor pathways. Importantly, the predictive accuracy in EPOS exceeded that of many later, more complex models, showing that simple bedside assessments remain clinically invaluable.

Algorithmic and modelling approaches have confirmed the primacy of trunk and leg control. The TWIST framework, tested in subacute stroke cohorts, identified trunk control and lower-limb strength as the strongest determinants of six-month walking independence (Stinear, Lang, Zeiler, & Byblow, 2017). This aligns directly with EPOS findings and reinforces that gross motor indicators capture the functional integrity of locomotor circuits more effectively than demographic or comorbidity factors alone.

Functional performance measures further refine prognostication. Multiple studies in the review demonstrated that endurance-based tests provide greater ecological relevance than speed alone. For instance, Selves, Stoquart, and Lejeune (2020) found the 6-minute walk test (6MWT) to be a stronger predictor of community walking than self-selected speed, highlighting that endurance reflects not only capacity but also the ability to sustain locomotion in real-world environments. Complementary evidence from Fulk, He, Boyne, and Dunning (2017) showed that thresholds of 0.49 m/s and 0.93 m/s distinguished between functional ambulation categories, yet it was the 6MWT that best discriminated step-count categories outside the clinic. This points to the value of adopting multiple outcome measures (see Section 2.7.3.1) to avoid over-reliance on a single metric.

Baseline cognitive and functional status also exert significant influence. In a large clinical cohort, Kwok, Potter, Dalton, and Myint (2012) showed that pre-stroke disability, measured using the modified Rankin Scale, was a strong predictor not only of gait outcome but also of mortality and length of hospital stay. Similarly, Wagle et al. (2011) demonstrated that models combining age, neurological impairment, and pre-stroke activities of daily living predicted long-term outcomes more accurately than motor impairment alone. These findings highlight that gait recovery trajectories emerge from an interaction of motor and non-motor determinants, and that cognitive and functional baselines must be incorporated into predictive models.

Biomechanical and kinetic predictors provide mechanistic depth. Evidence from propulsion-oriented studies demonstrates that kinetic measures outperform simple temporal-spatial outcomes in capturing underlying recovery potential. Bowden, Balasubramanian, Neptune, and Kautz (2006) showed that paretic propulsion scales closely with impairment severity and walking speed, and Awad, Reisman, Pohlig, and Binder-Macleod (2016) demonstrated that improvements in paretic propulsion during training explained a significant

proportion of variance in functional outcomes. These findings indicate that kinetic markers such as propulsion may serve as intermediate predictors linking impairment-level biomechanics with functional recovery. In parallel, Barroso et al. (2017) demonstrated that combining synergy features with kinematic and kinetic data predicted walking impairment with greater accuracy than the Fugl–Meyer scale, supporting the integration of mechanistic and clinical metrics.

Cross-analysis across the included studies identifies three converging insights:

1. Trunk control and early leg movement are indispensable: the EPOS and TWIST frameworks confirm their dominant predictive value.
2. Endurance and propulsion measures refine prediction: both the 6MWT and paretic propulsion explain aspects of recovery not captured by gait speed alone, providing ecological and mechanistic depth.
3. Broader functional and cognitive context matters: pre-stroke disability and baseline cognition shape trajectories, preventing over-simplification of recovery as a purely motor phenomenon.

Limitations of the evidence include a heavy reliance on acute/subacute cohorts with relatively standardised rehabilitation pathways, raising questions about generalisability to chronic stroke populations or to health systems with different service structures. Many predictive studies also used walking independence at six months as the principal outcome, which may not capture long-term adaptation or community participation. Furthermore, AFO use was often excluded, restricting applicability to real-world practice where orthotic support is common.

This review shows that independent gait recovery after stroke can be predicted with high accuracy using early motor control measures, particularly trunk

balance and voluntary leg movement. Predictive accuracy is enhanced when these indicators are complemented by endurance tests, propulsion measures, and baseline cognitive/functional status. Together, the evidence advocates for prognostic models that are both clinically simple and biomechanically informed, enabling clinicians to distinguish between patients likely to achieve true locomotor recovery and those who may remain limited despite apparent early gains.

2.7.4 Common Practices in Stroke Rehabilitation

Rehabilitation practices for post-stroke gait have diversified considerably in the past two decades, reflecting both advances in technology and a growing recognition of the multifactorial nature of recovery. While traditional physiotherapy remains the foundation of care, contemporary studies highlight that the effectiveness of rehabilitation depends not only on treatment type but also on its intensity, timing, and alignment with individual impairments.

Scoping and systematic reviews illustrate the breadth of interventions in current practice. A recent scoping review by Teodoro, Fernandes, Castro, and Fernandes (2024) identified thirteen categories of gait rehabilitation interventions, spanning conventional therapeutic exercise, treadmill-based locomotor training, functional electrical stimulation (FES), robotic-assisted gait training (RAGT), biofeedback, and non-traditional modalities such as Tai Chi and dual-task training.

The review underscored two themes:

1. Rehabilitation is increasingly multimodal, and
2. Technologies are not replacements but adjuncts to targeted therapy.

Importantly, while the review synthesised broad categories, it also highlighted persistent heterogeneity in dosing, outcome measures, and patient selection, complicating evidence synthesis and translation into routine care.

Evidence from targeted intervention trials emphasises the need to go beyond gait speed as discussed in section 2.7.3.1. Similarly, Awad, Reisman, Pohlig, and Binder-Macleod (2016) demonstrated that gains in paretic propulsion during training explained significant variance in functional improvement, again showing that mechanistic markers can add clinical value when combined with conventional measures (see Section 2.7.3.4).

Technology-assisted interventions illustrate both potential and challenges:

- Robotic-assisted gait training (RAGT): While devices allow high-intensity, repetitive practice, EMG evidence suggests that rigid assistance may reduce voluntary activation unless adaptive “assist-as-needed” protocols are used (Srivastava et al., 2016). Trials generally show improvements in walking speed and endurance, but results vary across devices and patient subgroups.
- Functional electrical stimulation (FES): Widely applied to improve dorsiflexion in swing phase, FES shows consistent short-term gains in clearance and gait safety. In some trials, such as Everaert, Thompson, Chong, and Stein (2010), long-term use was also associated with corticospinal plasticity, suggesting a role in true motor recovery rather than compensation (see Section 2.7.5). However, large multicentre RCTs (e.g., Kluding et al., 2013; Everaert et al., 2013) report that improvements in walking speed are broadly equivalent between FES and ankle-foot orthoses (AFOs), raising questions about ecological advantages.

- Biofeedback and hybrid treadmill paradigms: Biofeedback targeting tibialis anterior activity or trailing limb angle (TLA) shows promise for reshaping activation timing (Laufer et al., 2014), yet studies differ widely in thresholding methods and progression protocols, limiting reproducibility (see Section 2.7.5).

Common gaps across practices include limited integration with orthotic use. Multiple trials in the full-text corpus explicitly excluded participants wearing AFOs (e.g., Burpee & Lewek, 2015; Jonkers, Delp, & Patten, 2009), despite orthoses being a standard feature of post-stroke care (see Section 2.7.7). This exclusion raises concerns about external validity, as the studied populations may not reflect the “average” post-stroke patient encountered clinically. Moreover, findings from Burpee & Lewek (2015) caution that poorly tuned orthoses or fixed-timing FES may inadvertently block propulsion or induce over-correction of ankle kinematics. This highlights a translational gap: common rehabilitation practices risk being misaligned with real-world device use.

Cross-analysis of the included studies highlights three implications for practice:

1. Outcome selection matters: Reliance on SSWS alone is insufficient; endurance (6MWT) and biomechanical markers (propulsion, TLA) provide more nuanced indicators of recovery and should be standardised across trials.
2. Technology is not universally superior: While RAGT, FES, and biofeedback provide mechanistic and functional benefits, their effects are often comparable to conventional therapy when assessed by speed alone. Patient-centred outcomes (e.g., satisfaction, ecological transfer) therefore remain essential to judge clinical value.

3. Translation into practice is inconsistent: The frequent exclusion of AFO users, along with heterogeneous protocols and limited long-term follow-up, constrains generalisability. For rehabilitation to be truly evidence-based, protocols must reflect the devices and conditions that patients experience in clinical and community settings.

In summary, the evidence suggests that endurance and propulsion-related outcomes should guide practice more than speed alone, and that research designs must integrate real-world orthotic use to bridge the gap between laboratory findings and clinical realities.

2.7.5 Rehabilitation Approaches Targeting Gait and Muscle Activity

The rehabilitation of gait after stroke has moved beyond task repetition to embrace approaches that directly address neuromuscular deficits identified through EMG and biomechanical analyses. Across the reviewed studies, interventions that explicitly target propulsion, muscle activation timing, and limb alignment show the most consistent functional benefits, though their translation into everyday mobility remains uneven.

Propulsion-oriented treadmill training provides the clearest mechanistic pathway from impairment to recovery. Reisman, McLean, & Bastian (2010) showed that fast treadmill walking increased propulsion symmetry, and the addition of FES to plantarflexors in the FastFES paradigm further enhanced outcomes. In Awad, Reisman, Kesar, and Binder-Macleod (2014), FastFES produced clinically meaningful gains in speed (+0.18 m/s) and paretic propulsion (+3.28% of total propulsion), with retention at three months. Longer-term training yielded improvements in endurance, with responders in Awad, Reisman, Pohlig, and Binder-Macleod (2016) increasing 6MWT distance by almost 200 m on average and reducing energy cost by nearly 20%.

Crucially, modelling and simulation studies demonstrated that these gains were mediated by trailing limb angle (TLA) rather than ankle moment (Hsiao, Knarr, Higginson, & Binder-Macleod, 2015a; Knarr, Kesar, Reisman, Binder-Macleod, & Higginson, 2013), reinforcing TLA optimisation as the key driver of propulsion improvements (see Section 2.7.3.3).

Neurostimulation techniques such as FES extend this mechanistic rationale to dorsiflexion control and neuroplasticity. In a cohort of chronic stroke survivors, Everaert, Thompson, Chong, and Stein (2010) found that peroneal-nerve stimulation improved foot clearance and modulated corticospinal excitability, suggesting potential for motor recovery beyond compensation. Yet, when tested in large multicentre RCTs, FES performed no better than ankle-foot orthoses (AFOs) in terms of walking speed or endurance, with only user satisfaction tipping in its favour (Everaert et al., 2013; Kluding et al., 2013).

These discrepancies illustrate the persistent gap between laboratory markers of recovery and functional outcomes relevant to daily life.

Robotic-assisted gait training (RAGT) sits at the same intersection of promise and limitation. Devices guarantee high-intensity practice and safety, but EMG studies reveal that rigid assistance can suppress voluntary activation. Adaptive protocols encouraging patient effort yield better results, as Srivastava et al. (2016) demonstrated with improvements in tibialis anterior activity and interlimb coordination. However, when compared with conventional treadmill therapy, the additional benefit of robotics is often marginal, suggesting that the therapeutic context: intensity, progression, and patient engagement, may matter more than the device itself.

Biofeedback-based approaches offer a different route to targeting impairments. Visualising tibialis anterior activity during swing helped participants improve dorsiflexor recruitment in Laufer et al. (2014), while

propulsion-focused gait retraining studies demonstrated that improvements in trailing-limb angle and paretic propulsion were associated with more effective gait mechanics and enhanced walking performance (Hsiao, Knarr, Higginson, & Binder-Macleod, 2015a; Hsiao, Knarr, Pohlig, Higginson, & Binder-Macleod, 2015b).

Taken together, the evidence shows convergence on a small number of mechanistic themes. Across treadmill, FES, robotic, and biofeedback interventions, improvements in propulsion and endurance emerge most reliably when limb extension and activation timing are specifically targeted. At the same time, large-scale clinical trials remind us that gains in propulsion or EMG patterns do not guarantee superior walking performance or quality of life compared to simpler interventions. A further limitation is that many of these studies excluded AFO users (Burpee & Lewek, 2015; Jonkers, Delp, & Patten, 2009), restricting generalisability to real-world clinical populations where orthotic use is common.

In practice, therefore, the implication is not that advanced devices or stimulation are redundant, but that their true value lies in how well they integrate with established rehabilitation strategies and address specific biomechanical deficits. Future work must close the gap between laboratory-measured improvements and outcomes that matter for community participation, by embedding propulsion-oriented and EMG-informed protocols within multidisciplinary pathways and by designing trials that reflect the heterogeneity of real-world stroke rehabilitation.

2.7.6 Physiotherapy post-Stroke

Physiotherapy remains the backbone of stroke rehabilitation, with structured gait training forming a cornerstone of practice. The literature reviewed demonstrates that while conventional therapy consistently improves walking

speed, endurance, and balance, its impact depends heavily on training intensity, specificity, and alignment with underlying impairments.

Evidence for structured, task-specific training is robust. Studies have shown that treadmill-based interventions, particularly when delivered at faster-than-comfortable speeds, enhance gait symmetry and propulsion. Reisman, McLean, & Bastian (2010) demonstrated that fast treadmill walking reduced propulsion asymmetry, with later work extending these findings through the integration of FES (see Section 2.7.5). Gains in speed and endurance reported in these studies reinforce that high-intensity, task-oriented practice can drive both biomechanical and functional recovery.

Individualisation of therapy emerges as a recurring theme. EMG-informed approaches exemplify this. For instance, stiff-knee gait (SKG) may be caused by abnormal rectus femoris activity, excessive quadriceps co-contraction, or insufficient propulsion (see Section 2.7.2). Without EMG, these mechanisms are indistinguishable, and treatment risks being misdirected. Merlo and Campanini (2019) showed that dynamic EMG reduced inappropriate use of botulinum toxin injections to rectus femoris by 81%, highlighting the clinical value of tailoring interventions to specific neuromuscular patterns. This case illustrates how physiotherapy can be refined by integrating biomechanical evidence into treatment planning, rather than relying solely on observational assessment.

Technology-assisted physiotherapy complements traditional practice but brings mixed results. Laufer et al. (2014) found that surface EMG biofeedback improved tibialis anterior activation and dorsiflexion, while Srivastava et al. (2016) demonstrated that adaptive robotic training enhanced interlimb coordination compared with fixed assistance. Yet, larger RCTs involving peroneal-nerve stimulators or robotic devices reported outcomes equivalent to

simpler comparators such as AFOs or treadmill therapy alone (Everaert et al., 2013; Kluding et al., 2013).

These findings underscore the gap between mechanistic improvements, better EMG timing, more symmetrical propulsion, and functional outcomes like community walking or quality of life.

As seen in previous sections: persistent limitation of the physiotherapy literature is its exclusion of orthotic users. Several trials (Burpee & Lewek, 2015; Jonkers, Delp, & Patten, 2009) explicitly omitted participants using AFOs, even though such devices are widely prescribed in clinical practice. This design choice narrows internal validity but limits generalisability, leaving open questions about how physiotherapy and orthotic management should be combined in real-world rehabilitation. Without integration of orthotic effects, clinical translation of gait-training protocols remains incomplete.

The broader implication of this evidence is that physiotherapy should not be viewed as a monolithic intervention but as a framework for tailoring rehabilitation to patient-specific impairments. Intensive, task-specific practice improves walking performance, but its effects are most meaningful when guided by mechanistic markers such as paretic propulsion, trailing limb angle, and muscle activation timing. Technology-based adjuncts may augment therapy, but they do not consistently outperform simpler methods, and their role should be to complement - not replace - physiotherapist-led practice.

Finally, the omission of orthotic users in trials highlights a translational gap: effective rehabilitation in practice must be able to integrate physiotherapy, orthotic tuning, and impairment-specific strategies to ensure that laboratory gains transfer into meaningful improvements in community mobility.

2.8 Strength of Evidence and Risk of Bias

The structured appraisal process (Section 2.6.3.3) identified several methodological limitations that constrain the strength and interpretability of the current evidence base. While the volume of research in post-stroke gait is substantial, confidence in the findings is moderated by recurring risks of bias, variability in study design, and inconsistency in measurement approaches. The findings of this critical appraisal directly informed the interpretation of the literature presented in this chapter, ensuring that conclusions were drawn with appropriate consideration of study quality, methodological limitations, and risk of bias.

Across intervention studies, sample sizes were frequently small, limiting statistical power and increasing the likelihood of both Type I and Type II error. In addition, randomisation procedures and allocation concealment were often inadequately reported, particularly in quasi-experimental designs, introducing potential selection bias (Higgins et al., 2022). Blinding of participants and assessors was rarely feasible due to the nature of rehabilitation interventions, increasing the risk of performance and detection bias. These limitations were reflected in moderate-to-high risk of bias ratings across multiple domains using RoB 2 and ROBINS-I frameworks.

A further limitation relates to incomplete reporting and intervention heterogeneity. Key parameters, such as orthotic design characteristics, tuning procedures, or rehabilitation dosage, were frequently underreported or inconsistently defined. This reduced reproducibility and complicated interpretation of intervention effects. In orthotic studies specifically, failure to standardise or describe alignment variables introduced an additional layer of confounding, limiting confidence in attributing observed effects to the device itself.

Measurement bias was also evident. Differences in data acquisition and processing, particularly within EMG studies, introduced variability that was methodological rather than physiological. Inconsistent electrode placement, filtering, and normalisation procedures have been shown to significantly influence reported activation patterns, limiting comparability across studies (Baudet et al., 2014; Hug et al., 2021). Similarly, kinematic and kinetic outcomes were not uniformly defined or reported, further fragmenting the evidence base.

Finally, external validity was frequently constrained by selective sampling. Many studies excluded individuals with higher levels of impairment, comorbidities, or orthotic use, resulting in cohorts that did not reflect the heterogeneity of the stroke population encountered in clinical practice. This introduced a systematic bias towards higher-functioning individuals and limited the generalisability of findings to routine rehabilitation settings.

Taken together, these methodological limitations suggest that the current evidence base, while conceptually informative, should be interpreted with caution. The presence of consistent patterns across studies strengthens confidence in key mechanistic insights; however, variability in design, measurement, and reporting reduces certainty in the magnitude and transferability of observed effects.

These limitations reinforce the need for research approaches that prioritise methodological transparency, clinically relevant populations, and integrated outcome measures.

2.9 Conclusion and linking statement

The literature reviewed in this chapter demonstrates both the progress and the persistent shortcomings of stroke rehabilitation research and clinical practice. Substantial advances have been made in understanding post-stroke gait through biomechanical, kinematic, kinetic, and neuromuscular analyses. These studies move beyond the simplistic characterisation of hemiparetic gait as merely “slower” or “asymmetrical” (Section 2.3) and instead point to specific neuromechanical bottlenecks—most notably impaired plantarflexor timing, reduced paretic propulsion, and altered synergy modularity—that underpin functional limitations.

Importantly, while early clinical indicators such as trunk control and voluntary lower-limb movement remain strong predictors of eventual walking capacity, integrating biomechanical metrics such as propulsion, trailing limb angle, and synergy structure sharpens predictive accuracy and provides mechanistic targets for intervention (Sections 2.3 and 2.4). This underscores the need for outcome measures that go beyond speed or endurance to capture *how* walking is achieved and whether improvements reflect recovery or compensation.

Despite these insights, gait assessment in both research and clinical contexts remains dominated by temporal–spatial descriptors, particularly self-selected walking speed (Section 2.4.1). While convenient, these measures risk conflating true neuromuscular recovery with compensatory adaptations. Gains in speed, for example, can occur without improvements in propulsion or coordination, raising the danger of overstating functional recovery. This reliance on crude outcomes reflects not only methodological convention but also the absence of scalable, validated tools for measuring neuromechanical function in everyday settings.

This limitation is mirrored in orthotic practice. Although ankle–foot orthoses (AFOs) are widely prescribed in clinical care and recommended in guidelines for swing-phase clearance and stability (Section 2.5), most experimental protocols exclude orthotic users. The limited mechanistic studies available demonstrate that AFOs can meaningfully influence joint kinematics, muscle activation, and energy cost, but effects are highly sensitive to device design, stiffness, and shank-to-vertical angle (SVA) tuning.

Without systematically integrating orthotic use into rehabilitation trials, it remains unclear how these device-specific effects translate into long-term functional outcomes when combined with physiotherapy or gait retraining. This omission undermines the applicability of much of the research base to real-world practice.

High-fidelity gait analysis tools such as three-dimensional gait analysis (3DGA) provide unparalleled insights into joint kinetics, muscle timing, and synergy structure (Section 2.4.2). However, access is constrained by geography, resource intensity, and patient suitability (Section 2.4.3). For many survivors, fatigue, travel demands, and multi-hour protocols make 3DGA impractical. Clinicians are therefore often reliant on observational gait analysis or simple outcome measures, which lack the resolution to inform targeted, mechanism-driven interventions. This mismatch perpetuates inequities in access to advanced assessment and limits the ability to individualise rehabilitation.

Emerging wearable technologies offer a promising route to bridge this gap. Portable EMG systems, inertial sensors, sensorised insoles, and smart orthoses provide opportunities to capture propulsion, trailing limb angle, and muscle activation patterns outside the laboratory (Sections 2.4.3–2.4.5). If validated and integrated into service models, these tools could embed mechanistic insight into routine care, reducing reliance on crude descriptors

and improving the translation of research into practice. Yet their potential will only be realised if evidence demonstrates accuracy, acceptability, and clinical utility across the heterogeneous stroke population.

Taken together, the literature highlights the persistent tension between feasibility and mechanistic depth in post-stroke gait assessment, the unresolved debates around orthotic practice, and the promise, but as yet limited validation, of wearable systems. These gaps justify the direction of this thesis.

Accordingly, the following chapter sets out the research aims, hypotheses, and questions that guide this work. Specifically, this thesis seeks to:

1. Evaluate the potential of wearable EMG to characterise neuromuscular activity, with a particular focus on gastrocnemius function as a determinant of propulsion and gait quality.
2. Investigate the utility of the shank-to-vertical angle (SVA) captured through a low-cost and accessible system for a clinically feasible, tuneable measure for assessing and optimising gait in stroke survivors.
3. Explore how these measures can be integrated into orthotic practice to reconcile stability, propulsion, and efficiency in everyday rehabilitation.

In addressing these aims, the thesis sets out to advance the development of scalable, mechanistically informed, and clinically relevant outcome measures that bridge the gap between laboratory biomechanics and routine clinical rehabilitation. By doing so, it not only enriches scientific understanding of post-stroke gait but also contributes to more equitable and evidence-based rehabilitation practices.

Chapter 3 - Aims, hypotheses, research questions

As identified in Chapters 1 and 2: Stroke is one of the leading causes of long-term disability in the UK and worldwide, with gait impairment among the most common and enduring sequelae. Current rehabilitation practice is limited by a reliance on gait speed and observational analysis as primary outcome measures, both of which fail to capture the biomechanical and neuromuscular complexity of post-stroke walking (Awad et al., 2016; Kerr et al., 2020; Langhorne et al., 2009). Measures such as shank-to-vertical angle, trailing limb angle, and paretic propulsion have been proposed as more sensitive indicators of functional recovery, yet these remain underutilised in routine clinical settings and are rarely incorporated into outcome reporting. Additionally, many studies have indicated the importance of identifying muscle activity (EMG) in the leg when identifying potential for rehabilitation and qualifying effectiveness of post-Stroke gait.

At the same time, access to advanced three-dimensional gait analysis (3DGA) is scarce within the UK: services are unevenly distributed, resource intensive, and typically reserved for paediatric or highly complex neurological cases (Stewart et al., 2023). As a result, orthotic prescription and rehabilitation planning often proceed without detailed biomechanical evidence, potentially limiting patient outcomes and perpetuating inequities of access.

Recent advances in wearable sensor technologies—including inertial measurement units (IMUs), surface and textile-based electromyography (EMG), and activity monitors such as activPAL—offer an opportunity to address these gaps. Unlike laboratory-based systems, wearable devices have the potential to deliver clinically meaningful data at scale, both within clinics and in patients' natural environments. However, before such technologies can be integrated into practice, they must be rigorously developed, validated, and

benchmarked against existing gold standards to establish their accuracy, reliability, and usability.

The overarching purpose of this thesis is therefore to advance methods for the capture and interpretation of gait and muscle activity data post-stroke, with a particular focus on wearable technologies and their potential to inform rehabilitation pathways. The combination of wearable sensors for movement data and monitoring of EMG is still evolving in the field of Stroke rehabilitation, therefore the development and design of the final device identified through this thesis work is novel in its implementation.

This research is guided by three principal aims:

Aim 1 - To evaluate the current practices and infrastructure for gait assessment in UK stroke rehabilitation services.

This includes a critical analysis of current pathways, available facilities, and professional practices, including an original scoping questionnaire. The aim is to highlight where inequities exist in access to advanced gait assessment, to examine the extent to which current approaches rely on limited outcome measures such as gait speed, and to identify opportunities where novel technologies might enhance both assessment precision and service reach.

Aim 2 - To design and develop novel wearable technologies for the collection of gait and muscle activity data in post-stroke populations.

Building on limitations identified in the literature and initial scoping questionnaire, the thesis will report the iterative development of an innovative combination of a textile-based EMG “sock” and an Arduino-based Inertial Measurement Unit (IMU). The device is intended to overcome key barriers of

conventional systems: cost, accessibility, portability, and patient comfort. Development will be informed by clinical priorities (e.g., detection of foot drop, paretic propulsion) and by engineering requirements (e.g., signal stability, durability of textile electrodes).

Aim 3 - To validate these devices under controlled laboratory conditions.

An initial proof-of-concept cohort study and subsequent feasibility study will seek to validate the novel sensors against established gold standards such as Vicon motion capture and conventional Delsys surface EMG. The focus will be on assessing accuracy, repeatability, and robustness under test conditions with non-affected participants initially, before a limited trial with affected participants.

By establishing validity in laboratory settings, the work aims to provide a necessary foundation for subsequent application in clinical and real-world environments.

Taken together, these aims form a coherent programme of work. They begin with the identification of clinical and infrastructural limitations, proceed through the design and engineering of wearable technologies tailored to these needs, and culminate in laboratory-based validation on the intended user cohort. By linking clinical service realities with engineering innovation, the thesis provides both a critique of current practice and a pathway toward more equitable, data-rich, and personalised rehabilitation after stroke.

To demonstrate the alignment between the overarching aims of this thesis and the individual studies undertaken, Figure 5 maps each aim onto the corresponding methods, thesis chapters, and outputs.

This structure illustrates the coherence of the research programme: beginning with the identification of clinical and infrastructural limitations in current UK stroke rehabilitation practice, progressing through the design and development of novel wearable technologies, and culminating in laboratory validation against gold-standard systems.

Aim	Method / Approach	Thesis Chapter(s)	Outputs / Contribution
1. To evaluate the current practices and infrastructure for gait assessment in UK stroke rehabilitation services.	Nationwide questionnaire study; review of Sentinel Stroke National Audit Programme (SSNAP) data and clinical guidelines.	Chapter 5	Identifies disparities in access to 3D gait analysis; highlights over-reliance on gait speed; provides evidence of clinical need for alternative assessment approaches.
2. To design and develop novel wearable technologies for the collection of gait and muscle activity data in post-stroke populations.	Iterative engineering design and prototyping of textile EMG sock and Arduino IMU; integration with activPAL monitor.	Chapters 6, 7, 8	Establishes feasibility of low-cost, portable devices for post-stroke gait assessment; demonstrates design rationale grounded in clinical needs.
3. To validate these devices under controlled laboratory conditions.	Comparative validation against gold-standard Vicon 3D motion capture and conventional surface EMG with non-affected participants.	Chapters 7,8	Demonstrates accuracy, reliability, and comparability of novel sensors; provides foundation for future clinical and real-world application.

Figure 2 - Table indicating the three individual thesis Aims mapped against the Thesis chapters in which those aims are discussed and met.

Chapter 4 – Method Overview

This chapter presents the methodological framework that underpins the programme of research. It outlines the overall research strategy, philosophical positioning, and rationale for the staged study design. The intention is to provide the reader with a clear overview of how the studies were structured to address the thesis aims in a rigorous and systematic manner.

Consideration is given to the ontological and epistemological leanings of the researcher, as such reflection is integral to the scholarly rigour expected within a Doctor of Philosophy.

Finally, this chapter provides an overview of the individual study designs that collectively underpin the thesis. The rationale for the selection of each design is articulated, with attention to how methodological choices align with the overarching research questions and the philosophical positioning of the work.

4.1 Overarching study design

The included trials have a heavy bias toward quantitative methodologies.

The overarching study ‘strategy’ can be considered to be closest aligned with the concept of ‘Action research’, the four main pillars of which are; the research is of a practical nature, the research includes change as an integral aspect, a cyclical process is applied to the development of the research and finally, active participation practitioners within the research journey (Denscombe, M. 2010).

The cyclical process associated with the Action research strategy allows for integration of clinical experience, critical reflection of an issue, research (classically, systematic reviews), compilation of the research and planning of

intervention, action and finally circling back to integration into professional practice. This process is illustrated in Figure 3.

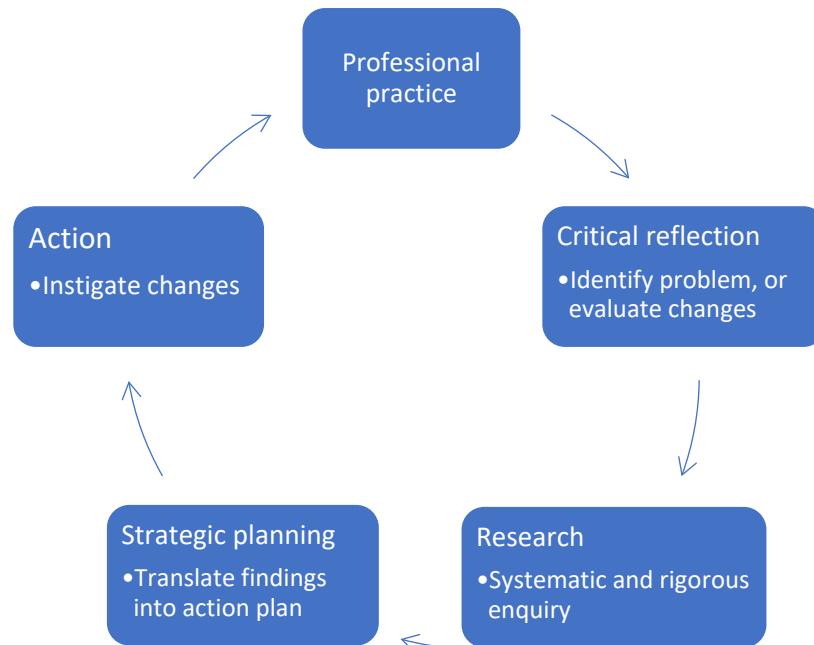


Figure 3 - Circular flow diagram to illustrate the Action research strategy through Professional Practice, Critical reflection, Research, Strategic Planning and Action back around to Professional Practice (Denscombe, M. 2010).

4.1.1 Key advantages of Action research:

- Participation of clinical practitioner – in the case of this research, the involvement between clinical Orthotist, biomedical engineers and industry partners.
- Practicality – addressing a clinical problem with positive practical outcomes that are better suited to be fed directly into practice.
- Continuous – a constant cycle of development and change is encouraged to help improve practices and resolve problems.

4.1.2 Known disadvantages with the research strategy:

- Limited scope and scale – this may limit assumptions made on the research outcomes.
- Impartiality – the Action researcher is unlikely to be detached or impartial to the research as the likelihood is that there is a vested interest in the outcome of the research. This is in contradiction to the historical viewpoint of ‘good’ scientific research (notably, quantitative research with a positivistic approach).
- Control – Due to the research settings and groups it may be more difficult to feasibly achieve a classic ‘control’ group when collecting data.

This piece of work therefore can be considered to be a blend of quantitative studies which are aligned with the Action research strategy in the stages 2 to 4 (as illustrated in Figure 6) wherein a clinical issue has been identified, systematic background research has been carried out and positive, practical solutions have been developed and tested in a reflective cycle. The outcomes of this research fall under the stages 4 to 5 (as illustrated in Figure 6) and thereby seek to provide a tool for actual practical change within a clinical setting.

4.2 Ontological and epistemological considerations

The classic Epistemological paradigm for bulk of this work is that of a Positivist paradigm; dealing with known facts and assuming there is a truth to be found. A term coined by Auguste Comte in the 19th century (Pomeroy et al., 2012). Furthermore, a deductive thinking process will be followed whereby a hypothesis will be formed from known theory, tested from observed results and either confirmed or rejected (formed from the Theory, Hypothesis, Observation, Confirmation model given in the ‘Research Methods Knowledge

Base' online textbook (Johnston, 2017)). This stipulation is met through the thesis Aims as set out in Chapter 3 - Aims, hypotheses, research questions. The individual studies which combine to create this piece of work each discuss separate aims with a similar overall theme.

On reflection however, the Positivist paradigm is limited. In line with the Action research strategy outlined previously, a Pragmatic paradigm would be more subtly suitable. Wherein a researcher views reality as containing elements that are available and independent of the conscious mind as well as elements that are created and therefore, dependent of the mind (Creswell & Clark, 2017). This differs from the black/white approach adopted with the Positivist paradigm but overall aligns well with research into rehabilitation. Pragmatist philosophy is strongly based on the concept that true knowledge and reality are based on the beliefs and habits of the individual and that these concepts are socially constructed (Yefimov, 2004).

Similarly, the lived experience of a person who has experienced a Stroke is unique and influenced by many factors, therefore, a finite conclusion cannot be drawn that will encompass all studied elements. The Biopsychosocial model of illness, first presented in 1977 by George Engel, describes the person as a combination of their biological, psychological and social factors and stresses that they, or their suffering, cannot therefore be seen as isolated 'organs' (Engel, G. L. 1977). This concept has been widely adopted in the social sciences domain, among others and demonstrates the importance of approaching treatments holistically (Wade & Halligan, 2017).

To link this approach with the Pragmatist theory in terms of Stroke, each person who experiences a Stroke will have a different lived experience up to that point (including, but not limited to, education, social and religious upbringing and pre-morbid health conditions). Following the Stroke event each person will experience unique combinations of effects (linked to the area of

tissue affected, time taken for treatment and method of intervention received as discussed in Chapter 1 – Introduction). Therefore, each participant involved in the research will have a different reality and concept of how they view themselves and ‘their’ Stroke. Similarly, the researcher will have their own unique perspective which will inevitably affect any descriptive analysis of the results.

However, the practicality of affecting research on humans will always have the issue of individual realities and the argument can be made that attempting to create a binary solution is therefore a flawed concept.

The philosophic framework to be followed therefore, will seek to maintain the balance between classic Ontological viewpoints of Direct realism and Subjective realism. Direct realism assumes of an objective scientific or systematic approach to the methodology and is favoured when the research question implies a ‘true or false’ objective outcome. This is in line with the explanation for ontological realism as given by Crotty that accepts “realities exist outside the mind” and can therefore be ‘known’ (cited in Burpee & Lewek, 2015). Subjective realism can be considered the polar opposite in that the researcher believes there cannot be a single explanation for a given phenomenon and seeks instead to explore the reality in which their work is situated (Ansari, Panhwar, & Mahesar, 2016).

As stated, the favoured approach for this work will include aspects from both Direct and Subjective realism as best suits the research design. The ontological and epistemological approaches guide the choice of study design to that of a quantitative biased approach. This is appropriate for research area and participant type. The scientific or systematic study design also requires the research to be conducted rigorously and methodologically.

4.3 Individual Study designs

As shown in Figure 4 this programme of work comprises three individual studies: an online questionnaire, a proof-of-concept cohort case study, and a feasibility study.

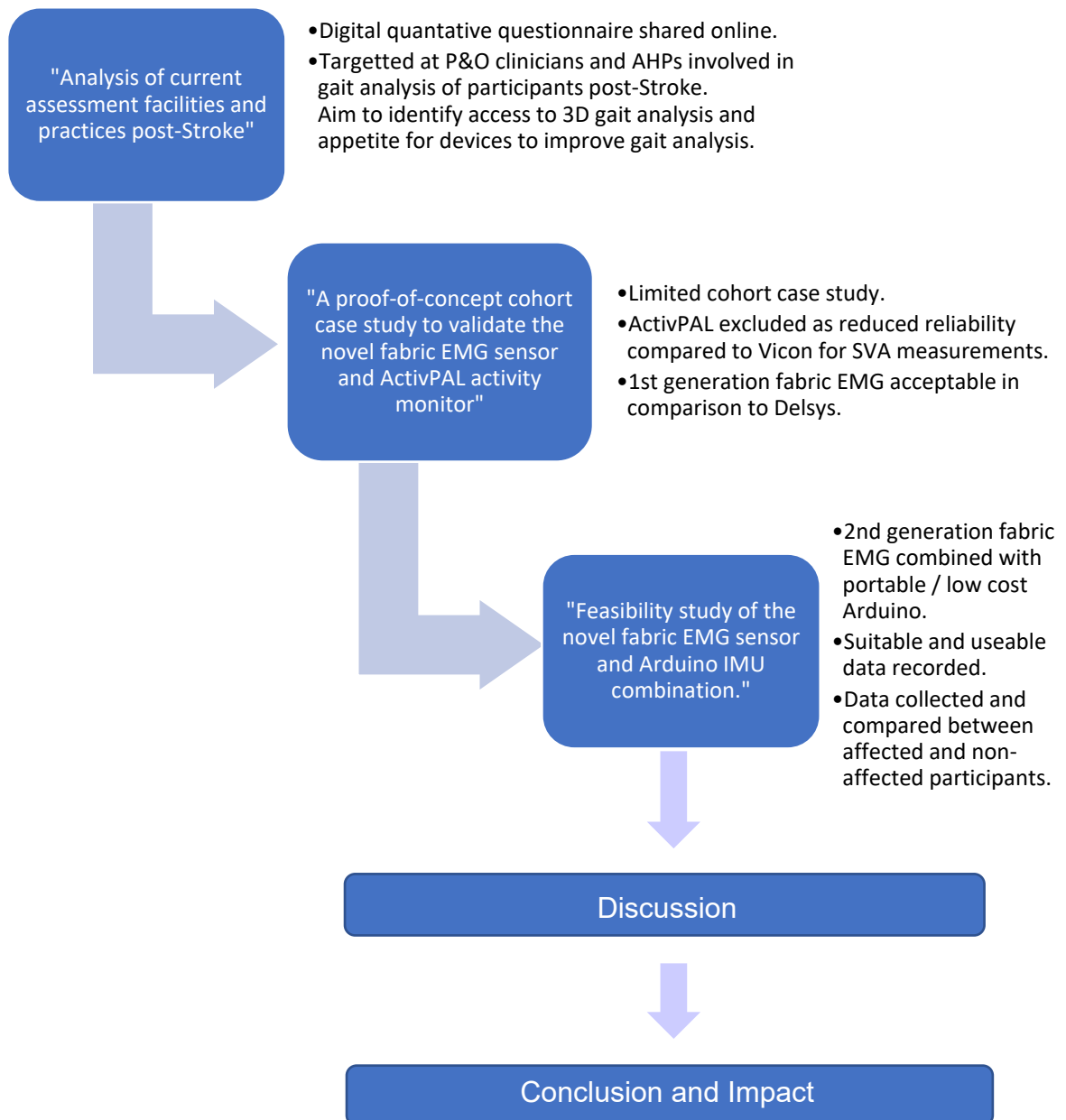


Figure 4 - Diagram illustrating the flow of the thesis through Studies 1, 2 and 3 into the Discussion and Conclusion Chapters. Additional notes highlight the study design for each individual section.

The figure also outlines how these studies connect, illustrating the progression of the research design and how each stage informs the next. Between studies 2 and 3 an iterative device design process took place with a small case study to test the device calibration and software coding before moving forward with the final testing phase. The flow of the thesis is represented through the sequential movement from individual studies into the later discussion and conclusion chapters.

As will be demonstrated, the findings from each study directly shaped both the design of the novel device and the methodological decisions underpinning subsequent phases of the research. The detailed process of device design, including the rationale for the initial development of a textile-based solution, is presented in Chapter 6.

4.3.1 Justification of study design choice

Selection of study designs was guided both by the developmental trajectory of the device (from concept testing to feasibility evaluation) and by the practical and ethical considerations of conducting research with vulnerable populations. As such, the methodological choices reflect best practice in medical device research, which emphasises incremental progression through proof-of-concept, feasibility, and early-phase evaluation before large-scale trials (Medical Research Council [MRC], 2006).

This staged approach not only ensured participant safety and compliance with institutional governance but also maximised the robustness of the evidence generated at each phase. The following subsections provide the justification for each of the three study designs undertaken: a questionnaire study, a proof-of-concept cohort study, and a feasibility study.

4.3.1.2 Justification for Questionnaire.

A questionnaire study was adopted at the outset of the research to capture clinician and end-user perspectives on the perceived needs, priorities, and barriers within post-stroke gait rehabilitation. Questionnaire methods are particularly well suited for gathering broad, structured insights from diverse stakeholder groups in a relatively short timeframe (Oppenheim, 2000). Within the context of device development, such insights play a critical role in ensuring that novel technologies are user-centred, clinically relevant, and aligned with real-world rehabilitation practices (van der Ploeg, 2016).

From a methodological standpoint, the use of a questionnaire provided several advantages at this early stage. First, it enabled the inclusion of a larger sample of stakeholders (relative to interview or focus-group methods), thereby capturing variability in clinical experience, patient demographics, and service contexts. Second, the structured format facilitated quantitative analysis of response patterns, while open-ended items allowed for qualitative elaboration where appropriate. This combination of structured and free-text data supports triangulation, enhancing both the breadth and depth of understanding (Creswell & Plano Clark, 2017).

In relation to the overall thesis, the questionnaire study provided critical foundational knowledge. By systematically identifying clinician priorities (e.g., safety, usability, accuracy of outcome measures) and clinician-reported patient concerns (e.g., comfort, ease of use, acceptability of wearing sensors), the study ensured that subsequent technical development and evaluation phases were grounded in user needs. This aligns with principles of co-design and participatory research, which are increasingly recognised as essential in the development of rehabilitation technologies (Greenhalgh et al., 2017).

Finally, given the resource constraints and external disruptions associated with the COVID-19 era, the questionnaire represented a pragmatic and safe method for collecting meaningful data without the need for face-to-face contact. Ethical approval was secured through the University of Strathclyde, and careful attention was paid to informed consent, data protection, and accessibility of the survey.

In summary, the questionnaire study provided an efficient and ethically sound means of generating user-centred evidence to guide the subsequent proof-of-concept and feasibility phases. It ensured that the development of the wearable system was not only technically robust but also clinically and socially responsive.

4.3.1.2 Justification for Proof-of-Concept Cohort Study during COVID 19 era

A proof-of-concept cohort study was selected as an appropriate design for the preliminary evaluation of a novel fabric EMG sensor in combination with an activPAL activity monitor. Proof-of-concept studies occupy a crucial role in medical device development, providing early evidence of technical performance and practical usability before progression to formal feasibility or clinical trials (Kistin & Silverstein, 2015). Such designs typically involve small, homogenous samples and controlled conditions, allowing researchers to refine protocols, identify methodological challenges, and ensure that devices perform as intended under laboratory settings (Leon, Davis, & Kraemer, 2011).

The restricted recruitment pool during the COVID-19 pandemic further necessitated a proof-of-concept approach. At the time of data collection, the University of Strathclyde operated under institutional COVID-19 safety protocols, which prohibited recruitment from outside the staff and student community and required strict adherence to physical distancing, hygiene, and use of personal protective measures (University of Strathclyde, 2020).

Consequently, participants were limited to healthy volunteers with campus access clearance, and each trial was conducted under conditions designed to minimise transmission risk. This was consistent with the wider context of UK higher education research, where universities implemented risk-managed approaches to sustain essential research while prioritising public health (Watermeyer et al., 2021).

Conducting a proof-of-concept study under these constraints ensured that preliminary data could still be generated despite external restrictions on recruitment and laboratory access. Importantly, the use of a homogenous, healthy cohort provided an opportunity to assess the integrity of EMG and IMU signals, validate activPAL outputs against gold-standard motion capture, and test the practicality of donning and doffing the device, all without exposing vulnerable clinical populations to additional risk during the pandemic period. Such staged testing aligns with best practice in early-phase medical technology development, in which safety, usability, and technical performance are prioritised before application to clinical populations (Medical Research Council [MRC], 2006).

In summary, the proof-of-concept cohort study design was justified both by the developmental stage of the wearable system and by the institutional and national restrictions imposed by the COVID-19 pandemic. By adhering to University of Strathclyde COVID-19 safety guidance, the study ensured that essential preliminary evaluation could proceed ethically and safely, laying the groundwork for subsequent feasibility testing in stroke-affected participants once wider recruitment was permitted.

4.3.1.3 Justification for feasibility study design

The adoption of a feasibility study design was considered appropriate for this research given the early stage of development of the novel wearable devices

and the need to establish methodological and practical viability before progressing to larger-scale clinical trials. Feasibility studies are widely recognised as an essential preliminary step in health research, particularly when testing innovative technologies or interventions in clinical populations (Bowen et al., 2009). Such studies aim not to determine efficacy but to assess key parameters such as recruitment, acceptability, safety, adherence, and the suitability of outcome measures (Eldridge et al., 2016).

In the context of post-stroke rehabilitation research, feasibility designs are especially relevant due to the heterogeneity of stroke populations in terms of age, severity, comorbidities, and functional limitations (Winstein et al., 2016). Conducting feasibility work allows researchers to evaluate whether protocols can be delivered safely, whether devices are acceptable to participants, and whether data quality is sufficient to justify subsequent clinical testing. Without this preliminary stage, there is a risk of progressing to costly and underpowered trials that may fail to recruit, retain, or generate usable data (Arain et al., 2010).

Furthermore, feasibility studies provide important effect size estimates that can inform sample size calculations for future trials (Lancaster et al., 2004). While statistical power is not the primary concern in feasibility research, exploratory analyses can help to identify meaningful outcome measures and guide methodological refinements (Thabane et al., 2010). In the current thesis, this is particularly relevant for validating the reliability of wearable EMG and IMU signals against gold-standard laboratory systems and assessing the practicality of using such devices with stroke survivors who present with variable motor impairment.

The use of a feasibility study design is therefore consistent with best practice in medical device development, where regulatory and ethical frameworks emphasise, staged evaluation beginning with safety, usability, and proof of

concept before larger-scale validation (Medical Research Council [MRC], 2006). By adopting this approach, the research ensures methodological rigour while protecting participant safety and optimising resource allocation.

4.4 Summary

This chapter has outlined the methodological framework underpinning the programme of research. Three interconnected studies were designed to build progressively: first, a national questionnaire survey to establish current clinical practice and service provision; second, a proof-of-concept cohort case study to generate preliminary data on novel wearable sensors; and third, a feasibility study extending this work to stroke-affected participants under enhanced ethical and safety protocols. Collectively, these studies ensured that the research advanced in a staged manner, moving from establishing clinical priorities, to testing technology in controlled conditions, and finally to assessing its application in a clinical population.

The logical progression of these studies provided both methodological rigour and practical relevance. Importantly, each stage directly informed the next: the questionnaire identified gaps and clinician needs, the case study provided early evidence of sensor viability, and the feasibility study demonstrated the practicality and safety of use with stroke survivors.

The next chapter focuses on the first of these studies, the national questionnaire survey. By examining current practice in post-stroke gait assessment and orthotic provision, the questionnaire provides the clinical justification for the development of a novel wearable device and establishes the foundation for the subsequent stages of research.

Chapter 5 - Analysis of current assessment facilities and practices post Stroke

5.1 Introduction

The debilitating and far-reaching effects of a Stroke have been extensively detailed in Chapter 1 and the current practices which focus on rehabilitation of walking (gait) post-Stroke have been explored through Chapter 2 – Literature Review. As has been previously discussed, despite its central role in rehabilitation, gait assessment remains dominated by subjective observation, with only a fraction of clinicians able to access advanced technologies such as three-dimensional gait analysis or electromyography (Baker, 2006; Stewart et al., 2023). National audits highlight ongoing inequities, particularly in orthotic prescription and tuning, where clinical decisions are often made without objective biomechanical evidence (SSNAP, 2022; Burpee & Lewek, 2015).

At the same time, emerging portable technologies present opportunities to transform practice by offering objective assessment in routine clinical settings (Picerno, 2017; Pomeroy et al., 2012). This chapter presents the results of a nationwide questionnaire survey of rehabilitation professionals, providing critical insight into current practices, barriers, and attitudes towards innovation in gait assessment.

The study aims are outlined in the following section.

5.2 Aim

Restoring gait has been identified as a central goal of rehabilitation. Although gait analysis can provide critical insights into walking performance and guide treatment, access to comprehensive assessment tools remains limited. Most clinicians rely on visual observation or video analysis, while gold-standard systems (e.g., 3D motion capture, force plates, EMG) are confined to specialist laboratories and are rarely available in routine clinical practice (Burpee & Lewek, 2015; Langhorne et al., 2018).

Emerging wearable technologies offer potential low-cost, portable alternatives. However, the successful translation of such technologies depends on understanding clinician needs, current practice, and perceived barriers.

Accordingly, this study focused on two primary aims:

- To investigate current gait assessment practices in stroke rehabilitation, including the tools and parameters clinicians use, and the limitations they encounter in different clinical settings. Establishing the current landscape of practice provides baseline evidence on how gait is evaluated, highlighting gaps between ideal and real-world assessment. This ensures that subsequent device development is informed by the realities of clinical workflows and constraints.
- To evaluate clinician perspectives on the need for and potential adoption of novel gait analysis technologies, particularly low-cost and portable devices usable with orthoses such as AFOs and KAFOs. By capturing clinician attitudes toward new technologies, the study addresses the translational challenge of aligning innovation with clinical

demand. This is essential for ensuring uptake, clinical utility, and long-term sustainability of new devices in stroke rehabilitation.

Together, these aims seek to identify the current gap between rehabilitation technology development and clinical practice needs in the UK, positioning this work within a co-design framework that prioritises end-user perspectives (Pomeroy et al., 2012).

5.3 Method

5.3.1 Study Design

A cross-sectional online questionnaire survey was conducted using Qualtrics (University of Strathclyde licence). This design was selected to capture practice patterns and attitudes across a diverse, multidisciplinary population of clinicians with stroke rehabilitation experience. Furthermore, online delivery ensured that participation was anonymous and voluntary.

The survey was open for six weeks (01 February - 16 March 2023) and ethical approval was obtained from the Department of Biomedical Engineering Ethics Committee, University of Strathclyde (Ref: DEC.BioMed.2022.329) (Appendix B.1).

To enhance the validity and usability of the questionnaire, the draft version was independently reviewed by a researcher with expertise in both survey methodology and the use of the survey software platform. This review focused on clarity of item wording, logical flow, response burden, and alignment of questions with the stated research aims. Minor revisions were made in response to feedback, including adjustments to terminology, formatting, and ordering of items. This process contributed to face and content validity by

ensuring that the instrument was comprehensible, relevant, and practically deployable within the intended participant groups.

A full copy of the Qualtrics questionnaire can be found in Appendix B.2.

For the context of this study 'Gait Analysis' was taken to refer to any observation of gait in the rehabilitating patient – regardless of method, intention or tools used.

5.3.2 Participants

Eligible participants were clinicians with at least six months' experience of working with stroke patients within the past five years. This included orthotists, physiotherapists, occupational therapists, nurse specialists, doctors, surgeons, and healthcare assistants. Inclusion criteria required an ability to read and comprehend English and access to an online device. Exclusion criteria included no recent experience (within the last five years) and inability to complete the questionnaire in English.

This broad inclusion reflected the multidisciplinary nature of stroke rehabilitation, where gait training may be addressed by professionals across different specialties (Pollock et al., 2014).

5.3.3 Recruitment

Recruitment followed multiple routes:

- Email invitations were circulated via professional and academic contacts.

- Social media campaigns (Twitter/LinkedIn) promoted the study with posts.
- Poster advertisements were distributed via professional organisations, with clear eligibility criteria and assurances of anonymity.

Participation was voluntary, with no incentives offered, reducing risk of coercion. The main motivation presented was contribution to research knowledge and the improvement of stroke rehabilitation technologies.

5.3.4 Consent and Information

In line with ethical requirements, participants were presented with a Participant Information Sheet (PIS) at the start of the survey, a copy of this PIS can be found in Appendix B.3. This outlined the study purpose, procedures, anonymity, and researcher contact details. Consent was obtained electronically:

“I confirm that I have read and understood the Participant Information Sheet... I consent to being a participant in this project”.

Participants who declined consent were automatically exited from the survey.

5.3.5 Survey Structure

The survey comprised four blocks with skip logic:

1. **Permission and Consent** – information sheet and consent form.

2. **Demographics** – profession, years of certification, clinical setting, and typical contact time with stroke patients.
3. **Gait Assessment** – whether gait analysis was performed, tools used (e.g., full gait lab, video, visual), parameters considered, and perceptions of access or limitations.
4. **Orthosis-Related Practice** – experience with AFOs/KAFOs, input into provision or tuning, gait parameters used in decisions, and limitations of gait analysis with orthoses.
5. **Attitudes to New Technology** – whether participants would adopt a device enabling easy gait analysis with and without orthoses.

Questions included closed (single/multiple choice), Likert-type, and open-text responses. Open-text responses allowed nuanced insights into barriers and perceptions. A full breakdown of the Questionnaire can be found in Appendix B.2.

5.3.6 Data Handling

All data were collected and stored in accordance with the University of Strathclyde Research Data Management and Handling Policy and the study's approved ethics protocol (Ref: DEC.BioMed.2022.329).

In line with the University's data management guidelines, raw data were exported from Qualtrics to a secure University of Strathclyde OneDrive account accessible only to the named investigators.

5.3.7 Data Analysis

5.3.7.1 Quantitative Analysis

Descriptive analysis was performed using the graphs, frequency tables, and summary statistics automatically generated by Qualtrics. These figures included counts, percentages, means, and standard deviations where applicable. Additional tabulations were checked against raw exported data for accuracy.

No advanced inferential statistics were undertaken due to the exploratory nature and small sample size of the study.

5.3.8 Aims-Method Alignment table

Table 1 - Aims / Method alignment table to demonstrate the expected outcomes mapped to individual aims for Study 1 – Questionnaire

Study Aim	Method Element	Expected Outcome
1. Investigate current gait assessment practices in stroke rehabilitation	Questionnaire sections on demographics, gait assessment practices, tools used, parameters considered, and access/limitations	Quantitative and qualitative data describing how clinicians currently assess gait, which measures are prioritised, and what barriers exist
2. Evaluate clinician perspectives on the need for and potential adoption of novel gait analysis technologies	Questionnaire sections on orthosis-related practice, limitations of gait analysis with orthoses, and attitudes to new devices.	Clinician-reported demand, acceptability, and conditions for adoption of novel, low-cost gait analysis devices

The data presented in Table 1 provides an overview of the individual study aims for the Questionnaire study mapped against the method elements and expected outcomes.

5.4 Results

5.4.1 Participant Characteristics

Twenty-four healthcare professionals responded to the questionnaire. The largest professional group was physiotherapists (46%, $n = 11$), followed by orthotists (29%, $n = 7$). Other respondents included nurse specialists (8%, $n = 2$), healthcare assistants (8%, $n = 2$), and a single occupational therapist (4%). One participant (4%) selected “Other,” describing their role as working in a “specialist stroke gym attached to a rehabilitation unit.” No respondents identified as doctors or surgeons.

Professional experience varied considerably as indicated in table 2. One-fifth ($n = 5$; 21%) had been qualified/working professionally for less than one year, 25% ($n = 6$) reported 1–4 years of experience, another 21% ($n = 5$) reported 5–9 years, and one-third ($n = 8$; 33%) reported more than 10 years in practice.

Overall, the sample was broadly distributed across experience levels, with no single category dominating; although the average coded category corresponded to 5–9 years’ experience, this does not reflect a central tendency but rather a spread across early-, mid-, and late-career clinicians.

This data is further broken down by profession in Figure 5 which indicates the responders with the most experience (from time qualified) were Physiotherapists while Orthotist responses were even across the first three time periods and only had one responder indicate a registration period of greater than 10 years.

Table 2 - Table showing the self-reported years qualified by all responders. Given in number (n=) and % delimited by less than 1 year, 1 to 4 years, 5 to 9 years and over 10 years.

Years Certified	Count (n)	Percentage (%)
<1 year	5	21
1–4 years	6	25
5–9 years	5	21
10+ years	8	33

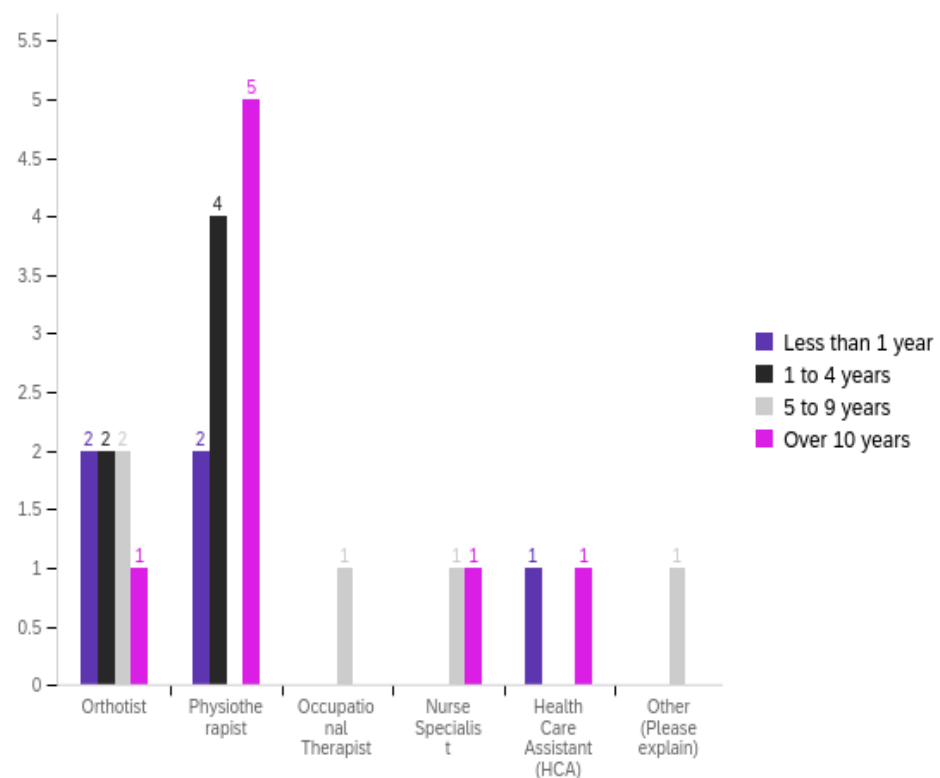


Figure 5 - Bar chart showing the self-reported number of years qualified of each responder (in years) separated into individual professions - Orthotist, Physiotherapist, Occupational Therapist, Nurse, Healthcare Assistant and Other.

Clinical settings were diverse, with multiple responses allowed. Most frequently, clinicians worked in hospital wards (30%) or community outpatient clinics (30%). Hospital outpatient clinics were selected by 20% of respondents, and home visits or care homes by 17%. Only 2% selected “Other,” again referring to a specialist stroke rehabilitation gym.

This information is shown in Figure 6 - Bar chart indicating the common settings where responders reported seeing / treating patients post-Stroke. The distribution highlights that gait assessment practices span both acute and community care contexts.

Q4 - In what setting do you normally see patients? (Multiple answers accepted)

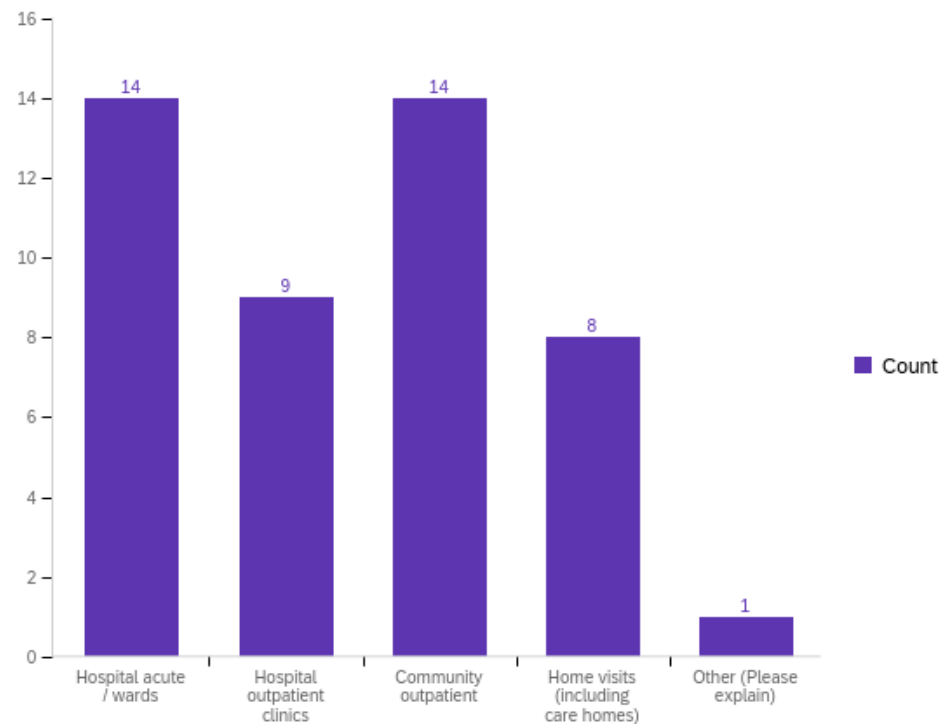


Figure 6 - Bar chart indicating the common settings where responders reported seeing / treating patients post-Stroke.

5.4.2 Stroke Patient Management

Almost all respondents (96%, $n = 23$) reported treating patients post-stroke, with only one respondent (4%) answering “No.”

Regarding contact time per session, the majority of respondents reported sessions lasting 10–30 minutes (65%, $n = 15$), with a further 30% ($n = 7$) reporting sessions longer than 30 minutes, and only 4% ($n = 1$) reporting sessions shorter than 10 minutes.

These findings suggest that most stroke patient interactions are time-limited, with relatively few opportunities for extended assessment.

Q5 - How much contact time do you normally have with each patient per session?

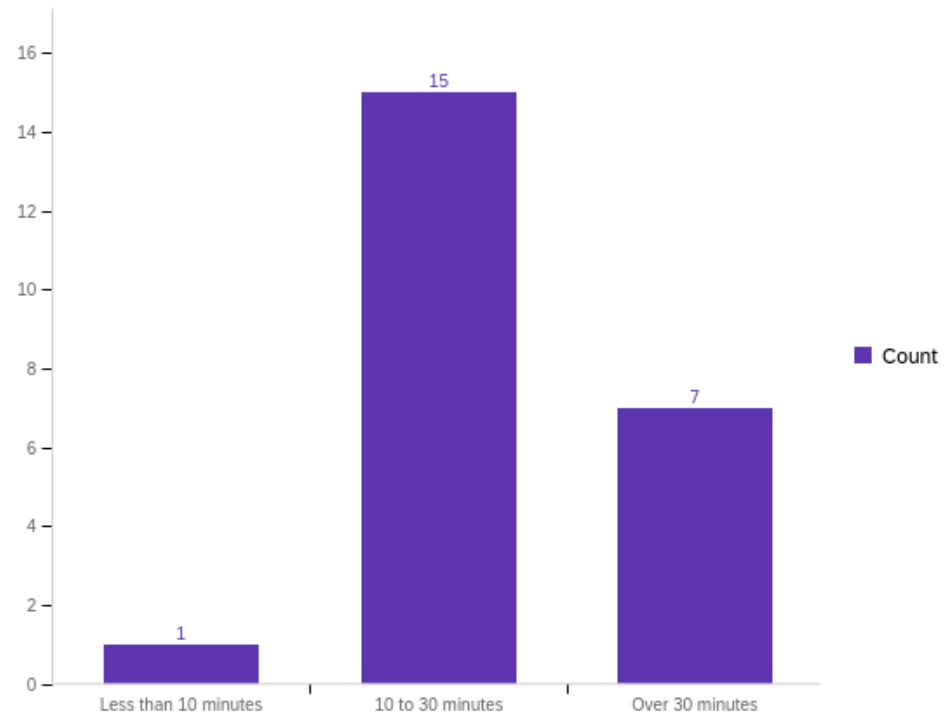


Figure 7 - Bar chart showing the spread of responses for average time spent with patients.
Under 30 minutes, 10 to 30 minutes and over 30 minutes

5.4.3 Use of Gait Assessment

Of 23 respondents, 87% ($n = 20$) reported routinely conducting gait assessments as part of treatment planning, as shown in Figure 8. A minority (13%, $n = 3$) reported not doing so, citing reasons such as lack of training, not being part of their role, or restricting observation to safety checks only.

In terms of tools used for gait assessment (multiple responses permitted, $n = 29$ responses total), the majority relied on visual observation ($n = 19$; 66%). Video analysis was reported by 28% ($n = 8$), and access to a full gait laboratory (3D motion capture, force plates) was reported by only 1 respondent (3%). An additional 3% noted occasional use of mobile phone video.

Q6 - Do you normally do a gait assessment as part of your treatment and planning...

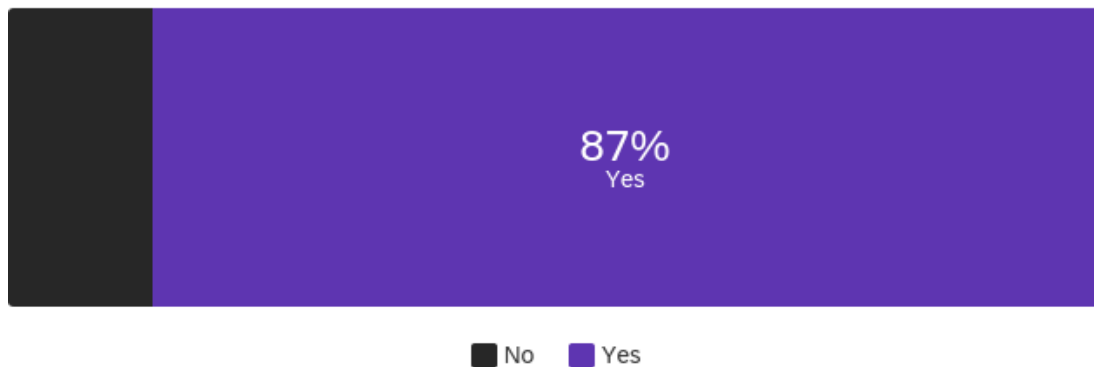


Figure 8 - Illustration showing Yes vs No responses for Question 6 relating to whether the respondent routinely performs a gait analysis as part of their assessment.

5.5.4 Parameters Considered in Gait Assessment

Respondents reported a range of gait features considered during assessment and treatment planning (multiple selections allowed, $n = 68$ responses total). The most common were gait speed ($n = 18$; 26%), step/stride length ($n = 16$; 24%), and joint range of motion ($n = 14$; 21%). Shank inclination was selected by $n = 11$ (16%), while muscle activity/EMG and limb power were rarely used ($n = 3$, 4%, and $n = 2$, 3% respectively). Only one participant (4%) stated that they did not use gait analysis in treatment planning.

This distribution suggests reliance on basic spatiotemporal and visual indicators, with limited integration of biomechanical or neuromuscular measures.

These data were further separated into individual professions which indicated high occurrences of the use of gait speed, step length and muscle ROM by Physiotherapists (who indicated a 100% use of the first two and 64% use of ROM in their profession-specific response) and Occupational Therapists ($n=100\%$, $n=0\%$ and $n= 100\%$) and Orthotists ($n= 86\%$, $n=86\%$ and $n=71\%$).

Q9 - Which aspects of the gait do you use during an assessment or when planning...

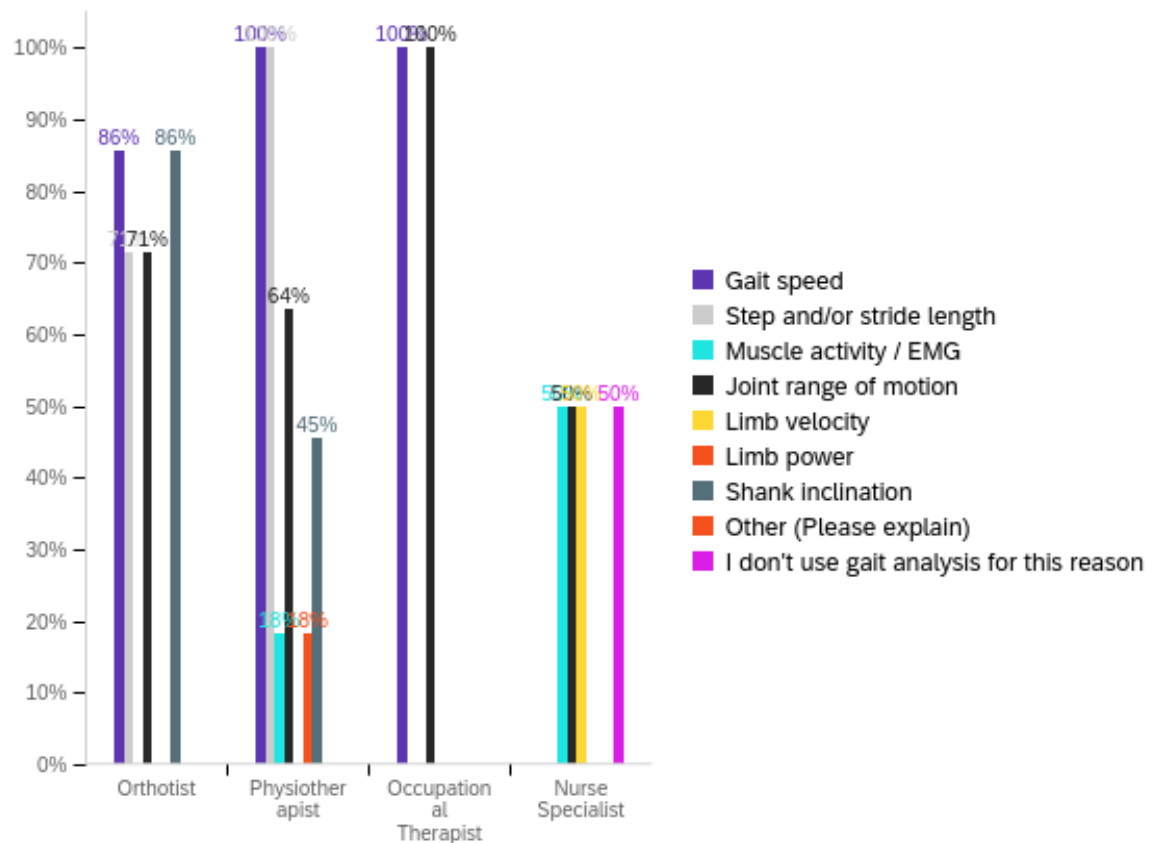


Figure 9 - Bar chart showing the breakdown of parameters of gait used by each profession (%)

5.4.5 Access to Gait Analysis

When asked whether they felt they had access to suitable gait analysis for every patient, most respondents (74%, $n = 17$) answered No, while only 26% ($n = 6$) answered Yes. This is clearly shown in Figure 10.

Among those reporting limitations (multiple responses permitted, $n = 34$ total), the most common barriers were lack of equipment (38%, $n = 13$) and limited

time with patients (32%, $n = 11$). Other responses included lack of training (9%, $n = 3$) and a wide range of “other” issues (21%, $n = 7$).

Free-text comments included:

- Lack of safe or appropriate space.
- Difficulty referring patients to specialist services.
- Restricted access to cameras or trained personnel.
- Reliance on colleagues for detailed analysis.

Q10 - Do you feel you currently have access to suitable gait analysis for every patient?

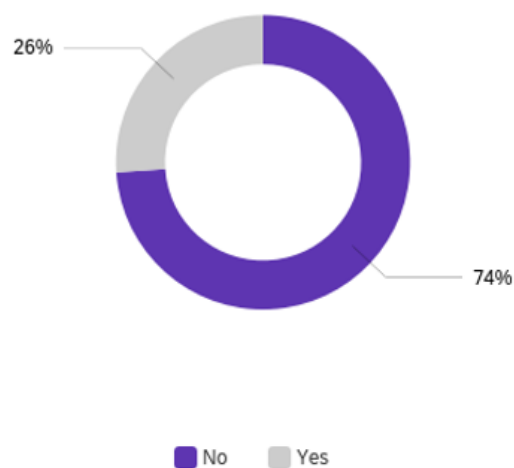


Figure 10 - Pie chart to illustrate the response to question 10 - Do you feel you currently have access to suitable gait analysis for every patient?

These findings underline structural and resource-related barriers to gait analysis in routine stroke rehabilitation.

5.4.6 Orthotic Practice and AFO Use

Most clinicians (91%, $n = 21$) reported treating patients who had been prescribed ankle-foot orthoses (AFOs), knee-ankle-foot orthoses (KAFOs), or knee orthoses post-stroke. Only 9% ($n = 2$) reported not working with orthotic devices.

Regarding input into orthosis provision/timing, 57% ($n = 12$) reported having direct involvement, 24% ($n = 5$) reported occasional or collaborative input (e.g., joint decisions with physiotherapists or orthotists), and 19% ($n = 4$) reported no involvement.

As shown in Figure 11: when asked what gait information was used when prescribing or tuning an AFO (multiple selections permitted, $n = 29$ responses), the most frequently considered factor was joint range of motion ($n = 12$; 41%). Other considerations included muscle activity/EMG ($n = 8$; 28%), shank inclination ($n = 3$; 10%), and gait speed ($n = 2$; 7%).

Several respondents provided additional free-text considerations such as patient safety, ground clearance in swing phase, foot position, and balance.

Q15 - What gait information do you use when prescribing or tuning an AFO? (Multiple answers accepted)

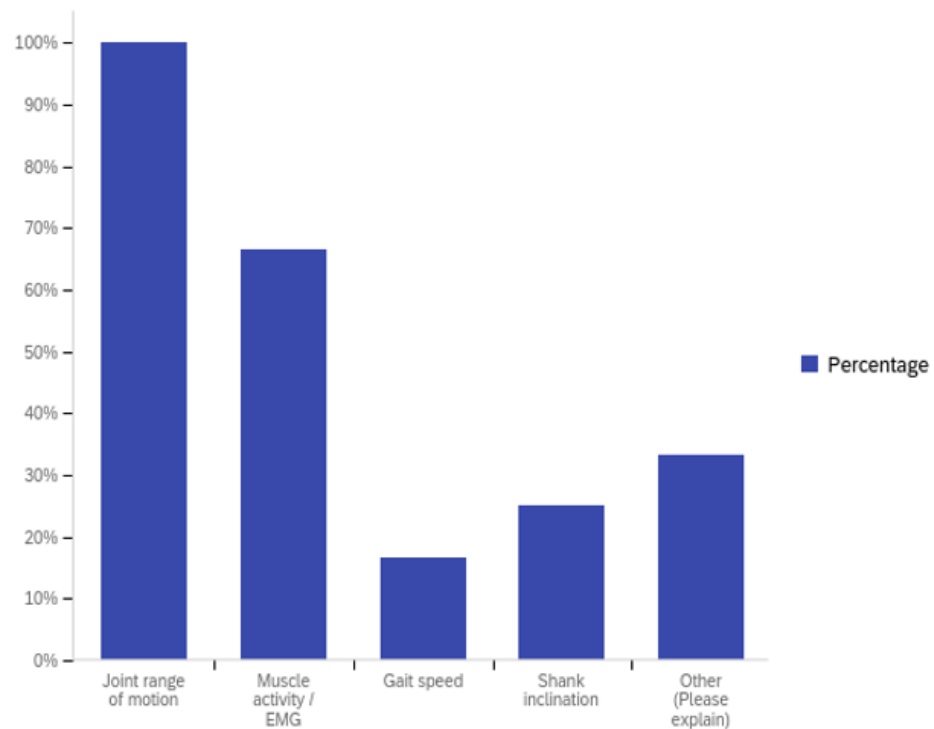


Figure 11 - Bar chart to show responses for question 15 - What gait information do you use when prescribing or tuning an AFO?

5.4.7 Limitations of Gait Analysis with Orthoses

When asked directly whether gait analysis was limited when patients wore orthoses, the majority (86%, $n = 18$) responded Yes, while only 14% ($n = 3$) responded No.

Qualitative explanations for perceived limitations included:

- Interference of AFO/KAFO with placement of motion capture markers or sensors.
- Splints masking underlying muscle activity.

- Altered gait mechanics making interpretation difficult.
- Added time burden in already short clinical sessions.

In contrast, those who answered “No” (14%) explained that visual observation was usually sufficient to determine whether the device was meeting clinical aims (e.g., maintaining safety).

5.4.8 Attitudes Toward New Devices

Q19 - If a device was available to help easily perform gait analysis on patients with or without an AFO do you think this is something you would use more regularly as part of your practice?

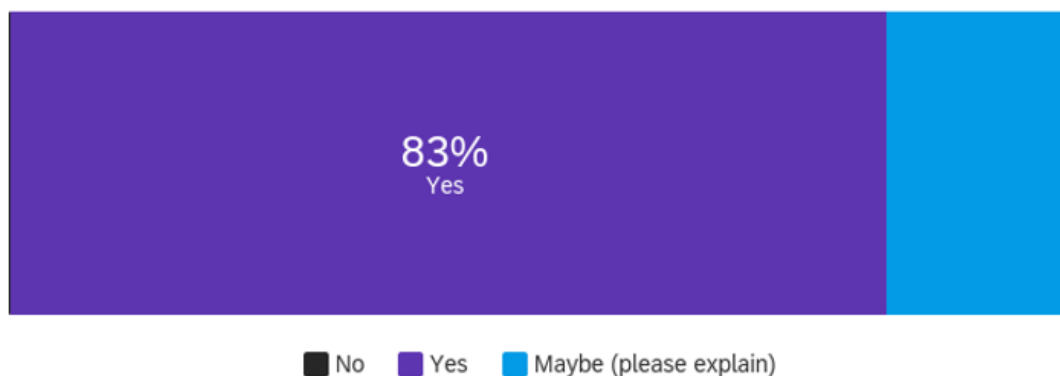


Figure 12 - Illustration showing 83% positive response when asked whether a device to help ease gait analysis on patients with and without and AFO would increase regularity of gait assessment as part of routine rehabilitation practice.

When asked whether they would use a device that allowed simple gait analysis for patients with or without an AFO, 83% ($n = 19$) answered Yes, and 17% ($n = 4$) answered Maybe, often qualifying their responses with the need for affordability, ease of use, and minimal time burden. No respondents answered “No.” This is illustrated in Figure 12

This strong positive response suggests a clear perceived demand for low-cost, practical gait analysis tools that integrate with orthotic practice.

5.5 Discussion

This questionnaire study provides an overview of current gait assessment practices in UK stroke rehabilitation and exposes critical gaps that directly influence patient care. The findings confirm that while gait assessment is widely undertaken, the methods used remain heavily reliant on subjective approaches, with very limited access to advanced technologies.

5.5.1 Reliance on Observational Methods

A key result was that 66% of clinicians reported relying primarily on visual observation for gait assessment. This finding aligns with Pollock et al. (2014), who reported similar reliance among UK physiotherapists, and with international data from the United States, where fewer than 15% of rehabilitation centres employed quantitative gait analysis (Damiano et al., 2013). Although observational assessment is simple and low cost, its reliability is limited: Rodriguez et al. (2019) found only moderate inter-rater agreement when rating hemiparetic gait, with subtle but clinically important deficits often missed. By contrast, quantitative gait analysis can detect impairments such as delayed gastrocnemius activation or reduced paretic propulsion (Boudarham et al., 2013), which are unlikely to be consistently identified through visual inspection alone.

Similarly, it was found that a very low (<5%) of respondents used EMG as part of their assessment strategy. However, almost 60% later stated that they observed 'muscle activity' during the fitting and tuning process for an Orthotic device.

This discrepancy suggests that, while clinicians recognise the importance of muscle function in clinical decision-making, they lack access to objective tools

to measure it. This finding is consistent with the literature presented in Chapter 2, which highlights a gap between the clinical importance of neuromuscular information and the availability of practical methods to quantify it in routine practice.

5.5.2 Limited Access to Specialist Gait Laboratories

Only 3% of respondents in our study reported access to gait laboratories. This reflects the structural inequity of advanced gait services in the UK, where laboratories are few in number, geographically concentrated, and often focused on paediatric or orthopaedic populations rather than stroke (Baker, 2006; Stewart et al., 2023). Comparable figures have been reported internationally, with van der Krogt et al. (2014) noting broader integration of gait labs into rehabilitation in the Netherlands, suggesting that service design rather than technological capacity drives access. The UK context therefore leaves most stroke survivors dependent on basic, subjective assessment, creating a postcode lottery of care quality.

5.5.3 Barriers to Detailed Assessment

The most frequently reported barriers were time constraints (72%), staffing shortages (54%), and cost (41%). These findings mirror national concerns, as physiotherapists, occupational therapists, and orthotists are listed on the UK Government's Shortage Occupation List (UK GOV, 2023). Similar barriers were reported in North America and Europe, where clinicians cited resource limitations as the greatest obstacle to implementing gait analysis (Damiano et al., 2013; Prenton et al., 2018). Notably, these results indicate that time pressures are perceived as a greater barrier than cost, suggesting that even relatively inexpensive technologies risk underutilisation if they cannot integrate seamlessly into existing, time-limited rehabilitation pathways.

5.5.4 Sample Size and Response Rate

One further limitation of the study is the small response size ($n = 24$). This constrains representativeness and raises questions about sampling coverage. On reflection, recruitment could have been enhanced through formal engagement with professional bodies such as the British Association of Prosthetics and Orthotics (BAPO) and national special interest groups (e.g., the UK Stroke Forum), which have established communication channels and could have facilitated broader dissemination across clinicians involved in stroke rehabilitation.

In addition, a more strategically targeted, multi-platform social media campaign, incorporating platforms such as Facebook and LinkedIn alongside Twitter/X, may have improved reach across different professional demographics, as evidence suggests that multimodal and professionally endorsed recruitment strategies are more effective in engaging healthcare populations in survey research (Cook et al., 2000; Topolovec-Vranic & Natarajan, 2016).

Nevertheless, context is important: the entire UK prosthetist/orthotist (P&O) workforce comprises only ~1,297 registered clinicians (HCPC, 2025), a small denominator compared with other allied health groups. Recent workforce studies place the number of employed clinicians even lower, around 926 (631 orthotists, 295 prosthetists) (Eddison et al., 2024). Within such a small profession, even modest samples can yield useful insights. However, meta-analyses of survey response trends suggest that web-only surveys typically achieve 20–30% response at best, often lower among clinicians, unless supported by incentives or formal professional endorsement (Daikeler et al., 2020; Nulty, 2008; VanGeest et al., 2007).

By that benchmark, the present study achieved below the “good practice” threshold from the P&O workforce, but this reflects the methodological realities of clinician recruitment.

Several factors likely contributed to the low uptake. First, the survey was distributed online without incentives, a method consistently shown to underperform compared to mixed-mode or incentivised designs (Cook et al., 2000). Second, workload and workforce shortages in stroke rehabilitation are well documented (BAPO, 2023; Stewart et al., 2023), leaving clinicians with limited capacity to engage with research. Third, recruitment relied on individual professional networks and limited social media, inherently missing those less digitally active. In addition, this approach relied heavily on existing communities of practice and randomly occurring cross-links between the professions targeted.

A further contributing factor may relate to professional identity and confidence within prosthetics and orthotics. Not all orthotists routinely manage stroke populations, and even among those who do, clinicians may be less likely to self-identify as specialists in stroke rehabilitation. Previous workforce and professional development studies have highlighted variability in role definition and confidence across the profession, which may reduce engagement with research perceived as condition-specific or outside core practice (Eddison et al., 2024). This may have led to under-representation of clinicians who contribute to stroke care but do not perceive themselves as part of a specialist stroke workforce.

As detailed previously, future changes to the search distribution would include seeking Professional body support for each profession targeted and broadening the list of platforms included.

Finally, survey fatigue is a growing challenge for clinicians, who are frequently targeted by institutional or research questionnaires. These barriers are not unique to this study but reflect systemic challenges in engaging small, pressured professional groups in research. While the small response limits generalisability, the patterns observed, reliance on observation, minimal access to gait labs, and enthusiasm for innovation, are consistent with larger surveys internationally, strengthening confidence in the validity of the conclusions.

5.5.5 Orthotic Prescription and Tuning

A particularly striking theme was the absence of objective methods for orthotic tuning. Respondents described difficulty in quantifying the effects of ankle–foot orthoses (AFOs), relying primarily on observation and patient feedback. This aligns with evidence that untuned orthoses can impair propulsion, exacerbate asymmetry, and increase energy cost (Burpee & Lewek, 2015; Mulroy et al., 2010). The absence of tools for objective orthotic optimisation represents a major unmet clinical need, with potential to limit rehabilitation outcomes if poorly configured devices restrict rather than support recovery. The consistency of this finding with both biomechanical literature and international surveys suggests an urgent need for innovation in this space.

5.5.6 Appetite for Innovation

Despite these barriers, the survey identified strong professional demand for new approaches: 82% of respondents reported willingness to adopt portable gait analysis tools, provided they were affordable, usable, and easily integrated into routine care. This finding is in line with Pomeroy et al. (2012), who emphasised that clinician usability and workflow fit are prerequisites for adoption, and with Picerno (2017), who argued that wearable systems offer

the most practical route to democratising gait analysis. The enthusiasm expressed by respondents suggests that the profession is not resistant to technology but simply lacks access to feasible solutions.

5.5.7 Critical Implications

The findings of this study highlight several interlinked issues with direct implications for stroke rehabilitation in the UK. The field remains over-reliant on subjective assessment methods, which cannot provide the objectivity required to detect subtle neuromuscular deficits or track longitudinal change reliably. There is structural inequity of access to advanced gait laboratories, leaving the vast majority of stroke survivors without the benefits of quantitative gait data. The systemic barriers identified—time and workforce shortages—suggest that future innovations must prioritise simplicity, speed, and integration into existing care pathways, rather than replicate laboratory-based methods. The study also revealed a profound gap in orthotic tuning; an area repeatedly identified in the literature as critical for functional outcomes but underserved by existing practice. Finally, clinicians expressed a strong appetite for innovation, with the majority prepared to adopt portable, low-cost tools if they can meet these requirements.

5.5.8 Synthesis with Published Literature

When compared with the wider literature, these results are consistent with international trends: clinicians worldwide rely on observation, have limited access to gait laboratories, and express willingness to adopt practical technologies (Damiano et al., 2013; Prenton et al., 2018). However, the present study adds novel, UK-specific evidence that orthotic assessment is a particularly urgent unmet need, and that clinicians perceive time rather than

cost as the dominant barrier. These insights are crucial in framing future innovation.

5.5.9 Link to Thesis Aims

These findings directly address Aim 1 of this thesis: to evaluate current practices and infrastructure for gait assessment in UK stroke rehabilitation. The results demonstrate that current assessment is dominated by subjective observation, that access to advanced gait laboratories is rare and inequitable, and that systemic barriers—particularly time and workforce pressures—limit the feasibility of detailed analysis. Importantly, the study highlights an urgent unmet need in the area of orthotic tuning and identifies strong clinician appetite for portable, objective technologies.

Together, these insights provide a robust evidence base that both clarifies the shortcomings of current practice and justifies the direction of subsequent chapters, which explore the potential of accessible, low-cost solutions to meet these clinical demands.

5.6 Summary

This chapter has demonstrated that gait assessment in UK stroke rehabilitation remains largely reliant on subjective clinical observation, with negligible access to advanced tools such as gait laboratories. Reported barriers were consistent across services, with time and staffing constraints outweighing financial considerations, and orthotic assessment identified as a particularly unmet clinical need. Clinicians highlighted the limitations of current approaches in objectively tuning ankle–foot orthoses, underscoring a gap between best practice guidance and feasible implementation in routine care.

Importantly, the survey also revealed a strong appetite for innovation. The majority of respondents expressed willingness to adopt portable, low-cost technologies capable of delivering objective gait data, provided these were practical, time-efficient, and directly relevant to clinical decision-making. This alignment between clinical need and openness to technological solutions provides a crucial mandate for innovation.

These findings directly inform the research decisions underpinning the next stage of this thesis. Chapter 6 builds on the gaps and opportunities identified here, describing the design and development of a novel wearable device intended to deliver objective, clinically meaningful gait data in a portable and accessible format.

The priorities articulated by clinicians – objectivity, efficiency, and integration into routine practice – have been embedded into the design specifications, ensuring that the device is shaped by real-world clinical constraints rather than theoretical ideals. In doing so, the trajectory from problem identification (Chapter 5) to solution development (Chapter 6) reflects the translational intent of this thesis: to move from characterising limitations in current practice to generating and testing pragmatic, evidence-based innovations.

Chapter 6 – Initial design and development protocol

6.1 Introduction

This chapter documents the design protocol and methodological choices which underpinned the initial developmental stages of the novel system.

The methodology follows a user-centred, engineering design process with rapid cycles of design → benchtop and human trials → feedback-led redesign.

Inspiration for this design protocol was taken from comparable doctoral work which has used similar iterative frameworks, for example, iterative hardware/control cycles in a bilateral arm exoskeleton for stroke rehabilitation (Gilson, 2023) and staged ergonomic evaluation of a robotic patient-transfer device with comparative testing against standard practice (Greenhalgh, 2020).

6.1.1 Identified Clinical Needs and Gaps

The literature review (Chapter 2) and clinician questionnaire study (Chapter 5) consistently highlighted critical shortcomings in current gait assessment and orthotic prescription practices.

These findings defined the clinical needs that the device was required to address:

i. Low-cost and accessibility

Advanced gait laboratories are rarely available in routine clinical practice, with only 3% of clinicians reporting access (Chapter 5).

A low-cost device was therefore essential to enable use beyond specialist centres.

ii. Portability and integration

With 72% of respondents citing time as a major barrier and 54% citing staffing limitations (Chapter 5), the device needed to be lightweight, portable, and simple to set up within routine clinical workflows.

iii. Clinically relevant outputs

Literature and clinician feedback identified gait speed, cadence, and shank-to-vertical angle (SVA) as the most meaningful outcome measures for gait assessment and orthotic tuning (Perry et al., 1995; Awad et al., 2016; Owen, 2004; Kerr et al., 2020; Tudor-Locke et al., 2018).

iv. Objective orthotic assessment:

Orthotic prescription and tuning were identified as unmet needs, with current practices reliant on subjective observation (Mulroy et al., 2010; Burpee & Lewek, 2015).

Objective measures were required to support more accurate ankle foot orthosis (AFO) alignment and an increase in person-centred care through personalised timing of intervention.

v. Suitability for stroke populations:

The target cohort required the device to be safe, tolerable, and comfortable, with particular consideration given to common post-stroke comorbidities such as reduced sensation or circulatory issues.

Together, these needs created the justification for developing a tool that could overcome current barriers, extend access to quantitative gait analysis, and provide clinically relevant data in everyday rehabilitation contexts.

The system has the potential to support more personalised orthotic and rehabilitation care by enabling decisions based on objective, individualised biomechanical data rather than subjective observation. Quantifiable measures of gait and muscle activity can inform more precise prescription and tuning of orthotic devices, including ankle–foot orthoses (AFOs), tailored to the individual.

Repeated measurement also enables tracking of change over time, supporting more appropriate timing and dosing of interventions. However, these benefits remain aspirational and depend on further development, validation, and integration into clinical practice.

Table 3 summarises the clinical gaps identified through the literature review (Chapter 2) and questionnaire study (Chapter 5), alongside the corresponding design requirements that informed the initial device concept. This mapping demonstrates how clinical priorities and practical constraints were translated into specific design features for the development of the novel device.

Table 3 - Mapping of Clinical Aims, Identified Gaps, and Device Requirements

Clinical Aim	Identified Clinical Gap	Corresponding Device Requirement
i. Low-cost and accessibility	Limited access to gait laboratories (only 3% of clinicians reported access)	Low-cost and accessible system deployable in community and hospital settings
ii. Portability and integration	Time and staffing constraints (72% and 54% of respondents, respectively)	Portable, rapid set-up device designed for integration into routine clinical workflows
iii. Clinically relevant outputs	Need for meaningful yet feasible outcome measures	Prioritisation of gait speed, cadence, and shank-to-vertical angle (SVA) as outputs
iv. Objective orthotic assessment	Reliance on subjective visual gait assessment (reported by 66% of clinicians) and lack of objective data for orthotic prescription and tuning	Provision of quantitative gait and muscle activity data to support evidence-based AFO alignment and personalised timing of intervention
v. Suitability for stroke populations	Requirement for comfort, tolerability, and safety in individuals with sensory or circulatory comorbidities	Safe, tolerable, and comfortable wearable design incorporating fabric-based EMG sensors

6.2 Initial Device Concept and Design Rationale

In response to these identified needs, the device concept was centred on creating a portable, low-cost system capable of simultaneously capturing muscle activity and gait kinematics. To achieve this, the design combined fabric-based EMG sensors, developed in collaboration with Footfalls and Heartbeats (Footfalls Smartex Limited, 2025), with an inertial measurement unit (IMU) to provide temporal and kinematic measures of gait.

The choice of gait speed, cadence, and SVA as primary outputs was directly informed by the literature and questionnaire findings, ensuring clinical relevance while remaining feasible for a low-cost system. The integration of EMG allowed for monitoring of muscle activation patterns, adding value to orthotic assessment where current practice relies heavily on subjective observation.

This conceptual framework provided the foundation for the iterative development process described in the following sections, in which successive prototypes were refined to balance signal fidelity, comfort, usability, and cost-effectiveness.

6.3 Focussed Scoping of the Literature

As several aspects required for the device design extended beyond the scope of the primary Literature Review presented in Chapter 2, a supplementary scoping review was undertaken. Whereas Chapter 2 focused on gait assessment and EMG analysis within stroke rehabilitation, this secondary search broadened the perspective to include evidence from adjacent fields. Specifically, the review explored literature on the use of compression garments and textile-based wearable sensors, drawing on both rehabilitation and non-clinical cohorts. It also examined industry guidelines regarding optimal sensor placement and investigated low-cost technological alternatives to fully equipped gait laboratories, recognising that affordability and clinical accessibility were central to the design objectives.

The focus of this additional review was directly informed by the findings of Chapter 2 and guided by the intended aims of the design exercise: to develop a feasible, low-cost, and clinically relevant monitoring tool. The outcomes of this targeted scoping review are presented in the following sections.

6.3.1 Scoping Review Methodology

The focussed overview presented in this section was undertaken as a scoping review rather than a full systematic review. The aim was not to exhaustively catalogue all available literature but to identify and synthesise key concepts from adjacent fields, compression garment biomechanics, textile-based EMG sensors, electrode placement standards, and low-cost gait analysis alternatives, that were directly relevant to the design of the novel device described in this thesis. This broader scope was necessary because these topics extend beyond conventional stroke rehabilitation research and were therefore not captured within the systematic review reported in Chapter 2.

6.3.1.1 Search strategy

Electronic searches were conducted across MEDLINE, Scopus, Web of Science, IEEE Xplore, and Google Scholar to capture both clinical and engineering literatures. The searches were performed between January 2004 and December 2024. Hand-searching of reference lists from relevant reviews and primary studies was also undertaken to ensure completeness.

Search terms combined concepts related to compression garments, electromyography (EMG), textile/wearable sensors, electrode placement, and low-cost gait analysis alternatives. Examples of search string combinations included:

“compression garment*” OR “graduated compression” OR “interface pressure”) AND (“biomechanics” OR “gait” OR “rehabilitation”)
 (“textile electrode*” OR “fabric sensor*” OR “wearable EMG”) AND (“signal quality” OR “impedance” OR “muscle activity”)
 (“electrode placement” OR “SENIAM” OR “sensor alignment”)

(“portable gait analysis” OR “low-cost gait” OR “wearable IMU” OR “sensorised insole”)

6.3.1.2 Inclusion and exclusion criteria

Inclusion criteria:

- Peer-reviewed articles, conference papers, or industry standards relating to compression garment biomechanics, textile electrodes, EMG sensing, electrode placement, or portable gait analysis technologies.
- Studies involving human participants (clinical or healthy populations).
- Publications from 2000 onwards, to reflect contemporary advances in wearable technologies.

Exclusion criteria:

- Studies not reporting empirical data (e.g., narrative reviews, opinion pieces).
- Non-English language publications where translation was not feasible.
- Paediatric or animal studies unless findings were directly transferable.
- Patents and unpublished technical reports (unless they informed background knowledge).

6.3.1.3 Quality appraisal and bias

Given the heterogeneity of the included evidence base (ranging from sports science to rehabilitation and textile engineering), a formal risk-of-bias tool such as RoB-2 was not appropriate. Instead, study quality was appraised using criteria adapted from Arksey & O'Malley's (2005) scoping review framework and the Joanna Briggs Institute (Peters et al., 2020): clarity of aims and

research questions, adequacy of methods (e.g., clear reporting of garment pressure measurement, electrode impedance testing), relevance of outcomes to biomechanics or EMG fidelity, transparency of reporting (sample size, population characteristics, limitations).

Where critical methodological details were missing (e.g., no description of garment pressure calibration or electrode-skin contact conditions), these limitations were noted in the synthesis.

6.3.1.4 Data extraction and synthesis

Data were charted according to the PICO framework where feasible:

Population (clinical vs healthy cohorts), Intervention/Exposure (type of garment, electrode, or device), Comparator (e.g., textile electrode vs Ag/AgCl electrode, compression vs no compression), Outcome (e.g., signal quality, impedance, kinematics, biomechanical effects).

Because of the diversity of study designs, a narrative synthesis was adopted. Findings were grouped thematically by domain (compression garments, textile electrodes, electrode placement, low-cost gait analysis), and key methodological strengths and weaknesses were highlighted. This approach ensured that the synthesis was both transparent and critical, while remaining flexible enough to accommodate engineering and sports science evidence alongside clinical rehabilitation studies.

6.3.2 Scoping Review Results and Discussion

6.3.2.1 Compression Garment Biomechanics and EMG Compression Garment Biomechanics

Compression garments (CGs) are widely used in medical and sports contexts, with proposed mechanisms including enhancement of venous return, attenuation of soft tissue oscillation, and modification of sensorimotor feedback (MacRae et al., 2011; Beliard et al., 2021). Despite theoretical benefits, systematic reviews highlight inconsistent effects on functional performance outcomes, with improvements more reliably observed in proprioception, comfort, and perceived stability than in objective kinematic or kinetic measures (Born et al., 2013; Engel et al., 2016). This discrepancy suggests that the biomechanical benefits of compression may be context-dependent and moderated by garment design variables.

A critical determinant of CG function is the magnitude and distribution of interface pressure. Graduated compression, in which pressure decreases from distal to proximal segments, is frequently recommended for both clinical and athletic applications. Clinical practice is generally constrained by established compression classification standards, such as the British Standard (BS 6612), which defines graduated compression classes (e.g., approximately 14–17 mmHg for Class I, 18–24 mmHg for Class II, and 25–35 mmHg for Class III) used routinely in UK prescribing and clinical practice (NICE, 2023).

However, *in vivo* pressure is highly sensitive to garment material properties, limb anthropometrics, and donning technique, with studies consistently reporting pressure decay over time and washing (Rimaud et al., 2016). These observations underscore the need for continuous verification of compression levels, particularly in longitudinal rehabilitation contexts where physiological changes alter limb volume.

While CGs reliably reduce muscle oscillations during dynamic activity (Doan et al., 2003; Beliard et al., 2021), the translation of this mechanical damping into clinically meaningful performance gains remains unresolved. Critically, the reliance on surrogate markers such as vibration attenuation may overstate the

functional importance of CGs unless corroborated by sustained improvements in gait mechanics or patient-reported outcomes. This tension highlights a gap between laboratory-measured biomechanical effects and real-world efficacy, an issue directly relevant to the present thesis in evaluating CGs as sensor platforms rather than performance enhancers.

Finally, it must be acknowledged that compression garments have also been associated with enhanced proprioceptive input through increased cutaneous stimulation and mechanical interaction with underlying tissues (MacRae et al., 2011). This effect can be conceptually linked to the principle of tensegrity, whereby the musculoskeletal system operates as a network of continuous tension and discontinuous compression, allowing external forces to influence internal structural alignment and sensory feedback. By applying distributed pressure, CGs may modulate this tension network, subtly enhancing joint position awareness and perceived stability.

However, while such sensory augmentation may lead to minor adaptations in movement strategy, it is unlikely to produce substantial changes in gait biomechanics in stroke-affected populations, where impairments are primarily driven by neuromuscular deficits rather than sensory input alone.

6.3.2.2 EMG Wearable Sensors (Textile Electrodes)

The integration of electromyography (EMG) sensing into textiles has gained traction as a route to unobtrusive, long-term monitoring of muscle activity. Textile electrodes fabricated from conductive yarns or coatings (e.g., silver-plated nylon, carbon inks, poly(3,4-ethylenedioxythiophene):polystyrene sulfonate (PEDOT:PSS)) can, under optimal conditions, yield signals comparable to standard Ag/AgCl gel electrodes while offering superior comfort and reusability (Kannaian et al., 2017; Taji et al., 2018). Nonetheless, electrode geometry and fabrication method critically affect impedance,

crosstalk, and durability, with no consensus yet on optimal design parameters across user populations.

A key limitation of textile EMG electrodes is their sensitivity to skin–electrode contact pressure. Evidence suggests a pressure “sweet spot” in the range of 15–20 mmHg, where impedance and motion artefact are minimized without compromising comfort (Ankhili et al., 2019; Finni et al., 2007). Pressures below 10 mmHg generally yield unstable signals, while excessive compression degrades both comfort and EMG fidelity. These findings align with the compression garment literature, suggesting that calibrated localised pressure is essential both for effective sensing and for wearability in rehabilitation scenarios.

Moisture management represents another challenge. Approaches such as incorporating hydrogels, moisture-retentive intermediates, or 3D knit structures have been proposed to stabilise impedance during long wear periods (Kannaian et al., 2017). However, these increase manufacturing complexity and may compromise durability, particularly under repeated washing. Oxidation of conductive fibres (e.g., copper) further reduces longevity, highlighting the need for protective coatings or alternative alloys (Taji et al., 2018).

6.3.2.3 Electrode Placement

Electrode placement is equally critical. International SENIAM guidelines provide standardised anatomical landmarks for surface EMG acquisition (Hermens et al., 2000), but garment-based integration must reconcile repeatable placement with ease of donning. Misalignment or drift during wear contributes to inter-trial variability and reduced signal reliability. Studies embedding conductive zones directly at SENIAM-recommended sites show

promise in reducing variability (Ankhili et al., 2019), but validation in clinical populations remains limited.

A single electrode site over the lateral head of gastrocnemius was selected to balance signal specificity with system simplicity within a wearable configuration. The lateral gastrocnemius is a key plantarflexor exhibiting consistent, phase-dependent activation during late stance and push-off, making it a reliable marker of gait cycle progression. Limiting acquisition to a single, superficial muscle belly also reduced susceptibility to motion artefact, electrode displacement, and inter-channel crosstalk, thereby improving signal stability during dynamic use. At this early stage of system development, this approach maximised the likelihood of obtaining robust and detectable EMG signals while minimising sources of signal degradation. This strategy aligns with the design objective of reducing setup complexity and facilitating rapid clinical deployment while retaining physiologically meaningful data.

Further justification for the selection of the lateral gastrocnemius is provided in Section 7.4.3.

6.3.2.4 Low-cost alternatives to 3D Gait analysis

Despite its capacity to provide gold-standard kinematic, kinetic, and EMG data, three-dimensional (3D) gait analysis remains effectively inaccessible for most stroke survivors in the UK. National coverage is sparse; limited to a small number of accredited centres clustered around tertiary hospitals and University research centres, while procedural demands require patients to undertake multi-hour assessments involving repeated walking trials (NICE, 2023; Stewart et al., 2023). Unsurprisingly, uptake is low: national outpatient data show that increasing travel time sharply reduces utilisation, a “distance-decay” effect disproportionately affecting patients with mobility, fatigue, or cognitive impairments (Kelly et al., 2016; NIHR, 2018). In practice, this means that

patients most likely to benefit from precise biomechanical evaluation are the least able to access it.

Findings from the questionnaire study (Chapter 5) confirm the scale of this accessibility gap. While 89% of clinicians routinely assess gait, the overwhelming majority rely on visual observation (66%), with only 3% reporting access to gait laboratory facilities. Barriers were widespread, including time (72%), staffing (54%), and cost (41%), which together limit the feasibility of advanced assessment in standard care pathways. Crucially, clinicians expressed a strong appetite for innovation: 82% reported willingness to adopt portable, low-cost gait analysis or EMG-based devices if these could be integrated into everyday clinical practice.

These findings, combined with the results from Chapter 5 - Questionnaire study - “Analysis of current assessment facilities and practices post Stroke” strongly indicate that there is both a structural deficit in service provision and a documented clinical appetite for pragmatic, real-world solutions. Wearable compression garments with integrated EMG electrodes cannot replicate the full fidelity of motion capture and force platform analysis, but they address precisely the barriers identified by clinicians—portability, repeatability, and ecological validity.

Evidence suggests that textile EMG electrodes, when coupled with calibrated compression, can achieve clinically comparable signal quality to Ag/AgCl electrodes while offering the comfort and durability required for daily rehabilitation use (Ankhili et al., 2019; Finni et al., 2007). Embedding these technologies into routine care would not simply extend monitoring capacity; it would shift assessment out of the laboratory and into the environments where stroke recovery unfolds.

Taken together, the limited reach of 3D gait analysis and the overwhelming clinical endorsement of wearable alternatives establish a strong rationale for the development of compression-integrated EMG systems. These devices respond not just to a technological possibility but to an explicitly stated clinical demand: accessible, patient-centred monitoring tools that bridge the gap between specialist infrastructure and real-world rehabilitation practice.

6.3.2.5 Summary

In summary, textile-based EMG electrodes offer clear potential for wearable rehabilitation monitoring but remain constrained by unresolved issues of electrode pressure, moisture management, and long-term durability. Their integration within compression garments presents both opportunities—consistent electrode placement and stable contact pressure, and challenges, particularly regarding the need to calibrate compression for both biomechanical and sensing functions.

For this thesis, these insights directly inform the design rationale for developing a compression sock with embedded EMG sensing, where garment biomechanics and sensor fidelity must be co-optimised.

Due to the novelty of the intended design there are no direct comparisons in current published literature.

6.4 Design Methodology Overview

6.4.1 Iterative Prototyping Framework

Instead of following a linear pathway from initial concept to final product, each prototype was approached as an iterative learning stage. This process allowed

for critical feedback on usability, signal fidelity, wearer comfort, and clinical feasibility to ensure alignment with the device needs as highlighted in section 6.1.1 Identified Clinical Needs and Gaps. The insights gained at each stage informed subsequent refinements, ensuring that the device developed responsively and adaptively to both empirical testing outcomes and end-user requirements.

This iterative approach reflects established principles of engineering design and human-centred development, providing a structured yet adaptable route toward the creation of a low-cost and clinically relevant monitoring tool.

The development pathway comprised three stages, these stages are detailed in Figure 13 below.

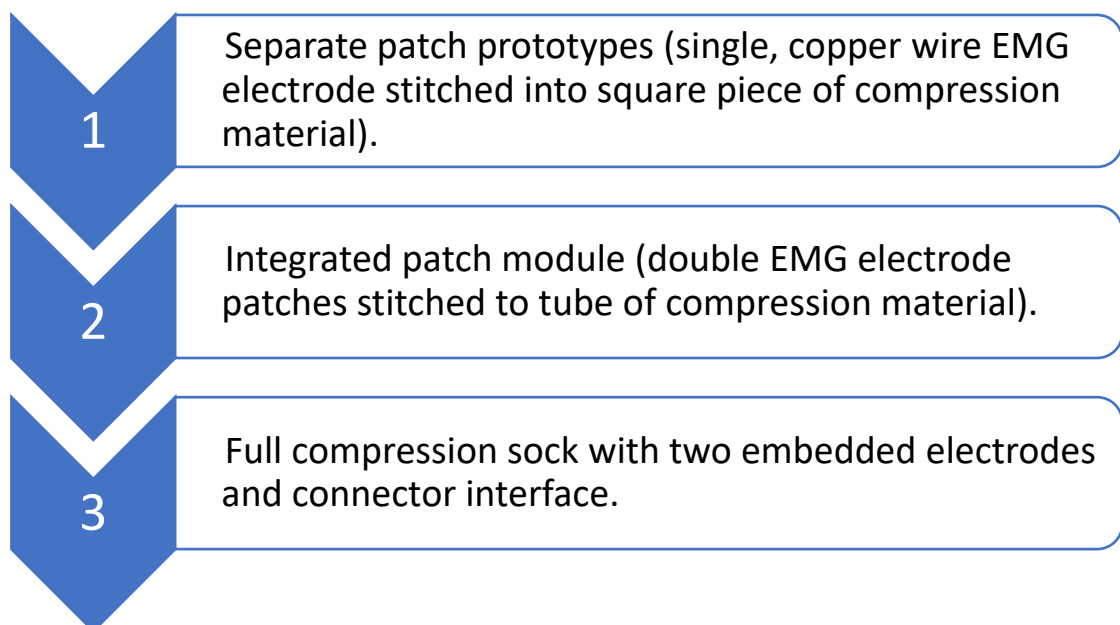


Figure 13 - Flow diagram which shows the three stages of Fabric electrode design from separate patches to a fully integrated sock.

An overview of the process is shown in Figure 14 below. These stages will be further detailed in sections 6.5 onwards.

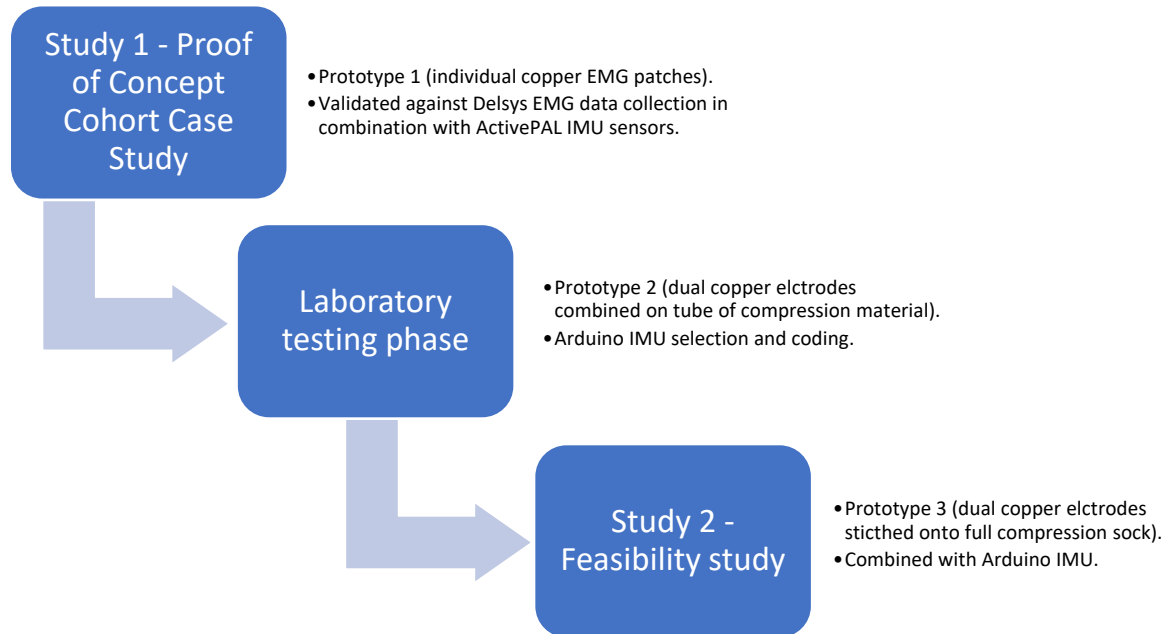


Figure 14 - Flow diagram to show the three stages involved in the device development.

As indicated above, two elements of the device design were tested and evolved through iterations 1, 2 and 3 (refer to Figure 13). Firstly, the fabric garment with integrated EMG sensor. Secondly, the IMU device which was used in collaboration with the EMG sensor and integrated with the fabric device.

While usability of the device is paramount to the overall success of the design process, comfort and wearability of the fabric element is also of high importance regarding the intended patient – people who have previously had a Stroke. This cohort is likely to have comorbidities affecting circulation and sensation and are thus at a higher risk of being negatively impacted through poor device design and fit. Consequently, a Human Centred Evaluation

protocol was observed during the development of the fabric element. This is explained further in section 6.4.2 below.

The development of the IMU component followed a staged process (Figure 15). In the initial phase, an ActivPAL4 device was utilised, as this system has been previously validated against the gold-standard Vicon motion capture system in the literature (Grant et al., 2008; Ryan et al., 2006). Its inclusion enabled the preliminary validation of the fabric-based EMG sensors under controlled conditions.

In subsequent stages, the IMU component was replaced with an Arduino-based device, selected primarily to address the requirement for a low-cost, accessible solution (refer to section 6.1.1 part *i*. Low-cost and accessibility).

The decision to adopt the Arduino platform was made during Stage 2, with integrated testing of the combined IMU and fabric EMG system undertaken in Stage 3 (Figure 18).

6.4.2 Human-Centred Evaluation

Testing of the fabric component at each stage prioritised wearability (comfort, heat, skin response), safety (compression class, contraindications), and data fidelity (stable electrode-skin contact during walking). Reflective practices were used through stages 1 and 2 (refer to Figure 13) and professional insight from the lead researcher, an HCPC qualified Orthotist. Semi-structured prompts enabled verbal user feedback during stage 3 data collection (refer to Figure 15).

This aligns with human-centred design practices from sensitized garment research, where comfort and stability are decisive for long-term adherence.

An illustration of the Human Centred Design process can be seen in Figure 15

- Illustration of the Human-centred evaluation loop — Showing inputs, prototype, test, feedback, iterate. (Health Foundation, 2024).



Figure 15 - Illustration of the Human-centred evaluation loop — Showing inputs, prototype, test, feedback, iterate. (Health Foundation, 2024)

6.5 Prototype Development Stages

As previously discussed through the flow diagram in Figure 13, the prototype was developed in three different stages. These will be discussed in detail below.

Note – this chapter considers the material choices and combinations.

Full device testing is aligned with the studies in subsequent chapters also incorporating statistical testing and validation against existing 3DGA and EMG sampling systems.

6.5.1 Stage 1 — Separate Patches

The initial stage of device development employed a hybrid configuration that combined validated commercial hardware with experimental textile-based electrodes. Specifically, a Delsys Trigno EMG sensor was connected to two separate fabric patches, each incorporating a copper-stitched electrode. The Delsys system was selected due to its reputation as a reference standard in surface electromyography (sEMG) research, with validation studies demonstrating excellent reliability in both controlled laboratory and clinical/functional contexts (Fuentes Del Toro et al., 2019; Hur et al., 2024). Data captured from the fabric electrodes via the Trigno system were synchronised and processed using Vicon Nexus, providing a high-fidelity reference for evaluation of signal quality (Vicon, 2016).

In parallel, an ActivPAL4 inertial measurement unit (IMU) was affixed to the participant to collect temporal gait parameters. Although no published studies have directly validated the activPAL for quantifying shank-to-vertical angle (SVA), the device's triaxial accelerometer allows estimation of segment orientation relative to gravity under quasi-static conditions. The activPAL has

previously demonstrated high reliability for detecting lower-limb inclination changes during postural transitions (Godfrey et al., 2015; Edwardson et al., 2017), supporting its sensitivity to segment-angle variation. Similar approaches using single-segment accelerometers have shown errors within 2–4° compared with optical motion capture (Leardini et al., 2014; Takeda et al., 2016).

Consequently, while the activPAL has not yet been validated specifically for SVA, its underlying sensing principles are consistent with those of devices successfully employed for this purpose. In this study, the activPAL was therefore repurposed as a shank-mounted inclinometer to estimate SVA, with the recognition that the device could represent a pragmatic, low-cost alternative to laboratory-based 3D motion capture. This testing phase allowed the additional validation of the activPAL against the 3dGA system Vicon. The experimental setup is shown in Figure 17.

A detailed description of the trial protocol and outcomes is provided in chapter 7.

Results from Stage 1 demonstrated that the fabric electrodes produced EMG signals broadly comparable to those acquired with traditional skin-mounted Trigno sensors, supporting their feasibility for muscle activity monitoring in dynamic conditions. However, several critical limitations were identified. First, accurate electrode placement posed challenges, as correct positioning of the fabric patches required alignment to anatomical landmarks following SENIAM guidelines (Hermens et al., 2000). This process was operator-dependent and introduced variability between trials. Second, inter-trial reliability was reduced due to the need to reposition each patch for successive recordings, increasing the potential for error and inconsistent electrode–skin contact.

Additionally, the use of separate fabric patches without integrated fixation led to substantial motion artefact and skin movement artefact, requiring extensive filtering during data processing. Such artefacts are a recognised limitation of surface EMG (De Luca et al., 2010), but in this configuration they were exacerbated by variability in patch adherence. Importantly, these issues conflicted with the predefined device aims of portability, accessibility, and ease of clinical integration (see Section 6.1.1).

Beyond technical performance, the reliance on the Delsys Trigno system and Vicon Nexus software presented barriers to real-world clinical use. Although the Trigno system is semi-portable, it remains costly and requires specialist expertise, while Vicon Nexus is a laboratory-based solution that demands dedicated training, software licensing, and infrastructure (Vicon, 2016). Such requirements are incompatible with the identified clinical needs for a low-cost, portable system deployable outside of gait laboratories (Chapter 5).

Collectively, these findings demonstrated proof-of-concept validity for the fabric electrodes but highlighted several operational and feasibility challenges. Consequently, a redesign phase was initiated to address these shortcomings, culminating in the development of Prototype 2.

6.5.2 Stage 2 — Integrated Patch Module

Building on the limitations identified in Stage 1 (see Section 6.5.1) and findings from Study 1 (detailed in Chapter 7), the second prototype was designed as an integrated tubular module. By utilising a circular-knitted polyamide/elastane fabric base (see Section 6.5.1), the prototype harnessed the compressive properties of the textile to improve donning and retention, thereby reducing motion artefact and skin movement noise while maintaining comfort for the wearer. Compression garments have been shown to enhance stability and consistency of sensor placement in wearable devices (Yin et al., 2015),

aligning with the project's aim to develop a clinically usable system for stroke populations.



Figure 16 - Image depicting the tube iteration of Prototype 2. With copper stitched EMG electrodes and button connectors visible.

As seen in Figure 16 two copper-stitched electrodes were integrated directly into the fabric tube. The fabric base with stitched electrode devices were supplied by the industry partner Footfalls and Heartbeats (Footfalls Smartex Limited, 2025). Each electrode measured 25 mm in diameter and was positioned with a 10 mm inter-electrode spacing to reduce crosstalk while approximating SENIAM-recommended placement distances (Hermens et al., 2000). Each electrode was supported by a corresponding attachment point located approximately 1 inch proximal to facilitate reliable connection to the EMG hardware.

To address the issues of cost and portability that remained unresolved in Prototype 1, the EMG hardware was replaced with a MyoWare SEN-13723 board, integrated in combination with an Arduino Mini IMU. This novel pairing allowed synchronous capture of both muscle activity and kinematic data within a single system, rather than relying on separate platforms as in Stage 1.

The interface and coding framework were initially developed with support from Prof. Philip Rowe (University of Strathclyde, BME Department) and subsequently adapted by the author during iterative testing.

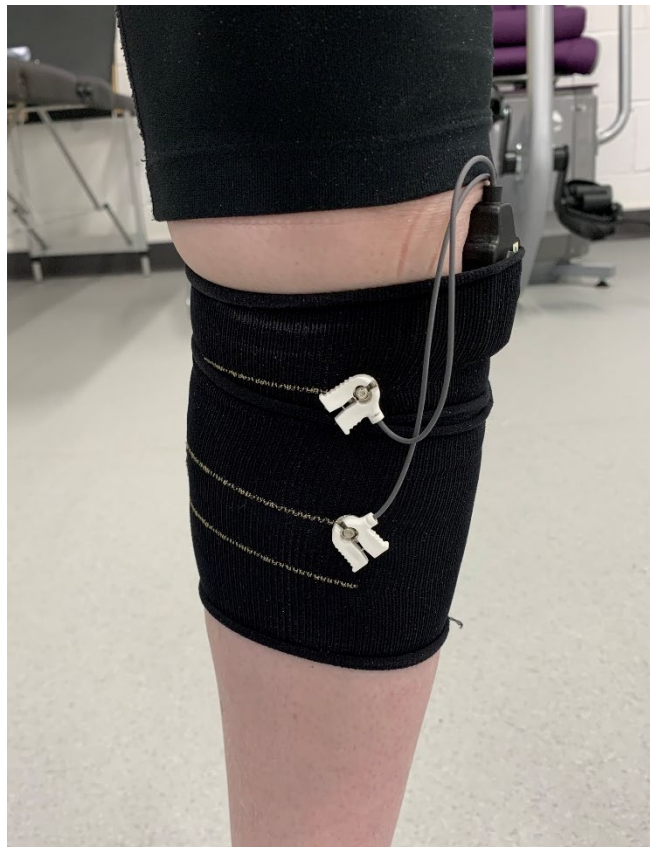


*Figure 17 - Coronal (anterior to posterior) view of integrated tube design during testing.
Delsys unit and snap leads visible anteriolaterally.*

Laboratory evaluation of this configuration, however, identified new challenges. While the EMG and IMU integration demonstrated functional

feasibility, the system was unable to reliably detect key gait events, namely initial contact (heel strike) and stance-to-swing transitions. To overcome this limitation, a Phidgets FlexiForce sensor (100 lb capacity) was incorporated into the design and coding pipeline. Force-sensitive resistors are widely used for low-cost gait event detection (Phidgets, 2017; Winter, 2009), and this addition significantly enhanced the system's capacity to delineate individual gait cycles.

Despite these improvements, repeated wear testing revealed limitations in the tube-based fabric design. Specifically, during prolonged use the material exhibited a tendency to shift and bunch proximally and distally, which increased localised pressure at the fabric margins. This not only reduced wearer comfort, a critical factor for stroke-affected populations with sensory or circulatory comorbidities (see Section 6.1.1) but also risked contact between copper-stitched electrodes, generating crosstalk and compromising data fidelity. These outcomes were inconsistent with the project's safety and usability aims.

In response, the fabric design was further refined to incorporate a commercially manufactured, ISO-standard Class 1 compression sock with incorporated stitched electrodes, produced by Footfalls and Heartbeats (Footfalls Smartex Limited, 2025). This provided a more consistent, clinically validated base garment, with durability and wearability characteristics aligned to medical compression standards. The development and testing of this improved design are described in Section 6.5.3.

The collaboration with the industry partner Footfalls and Heartbeats arose through their ongoing support of the Centre for Doctoral Training in Prosthetics and Orthotics, including the provision of materials and technical input. Initial introduction was facilitated by the PhD supervisor, who had an established collaborative relationship with the company. Footfalls and Heartbeats expressed a clear interest in supporting the project, particularly in relation to

the future development of integrated monitoring systems within textile-based garments.

6.5.3 Stage 3 — Full Sock Prototype

Informed by operational challenges encountered in Stage 2, the final prototype transitioned to a medically compliant graduated compression sock format. The electrode design was refined: each copper-stitched electrode was reduced to a 15 mm diameter, with an inter-electrode spacing of approximately 8 mm, and a stable 10 mm separation between opposing electrodes, aimed at minimizing inter-electrode crosstalk while adhering to SENIAM-recommended placement for the gastrocnemius lateralis (Hermens et al., 2000).

Electrode height and orientation on the donned sock were determined using standardised body-size reference data, targeting an adult cohort in line with the intended user cohort, corresponding to commercial compression sock sizes 3–4 and 4–5. For instance, ankle circumferences in the relevant range span approximately 21–26 cm, with corresponding calf circumferences of approximately 30–43 cm (based on intermediate size data from general compression sock guidelines)

Two sock sizes were produced—one covering mid-range adult dimensions (size 3–4: approx. 21–26 cm ankle, 30–43 cm calf) and a second covering larger adult dimensions (size 4–5: approx. 25–31 cm ankle, 35–48 cm calf)—to accommodate anthropometric variability in Stage 3 testing.

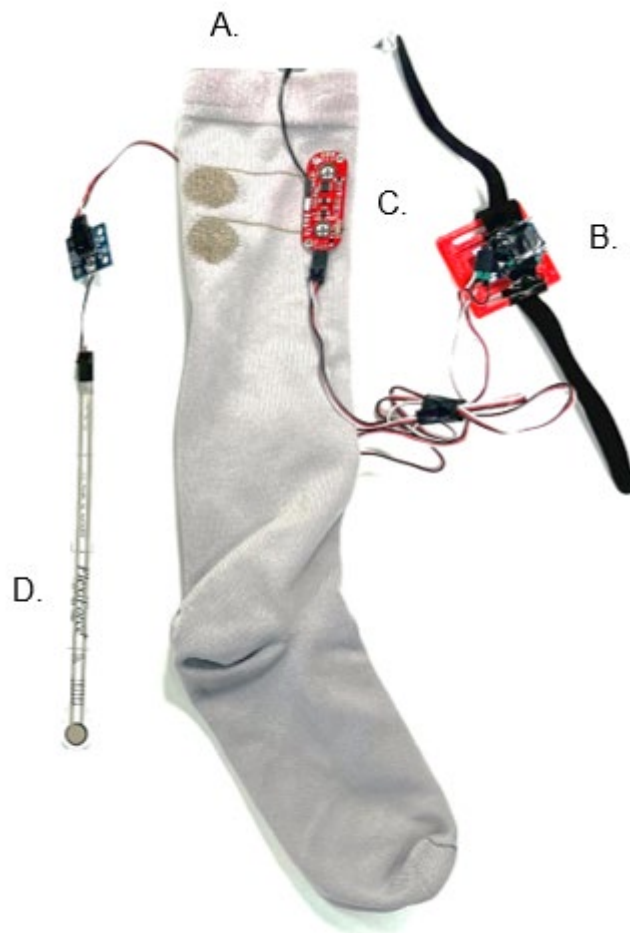


Figure 18 - Final prototype configuration integrating the compression sock with embedded stitched electrodes (A), MyoWare EMG board (B), Arduino Pro Mini housed in a 3D-printed anterior shank casing (C), and Phidgets FlexiForce sensor (D) positioned beneath heel.

Connectivity was streamlined: electrodes were stitched to enable low-profile snap-on connectors that interfaced directly with the MyoWare Muscle Sensor (SEN-13723) EMG board (Advancer Technologies, LLC, n.d.), while thin, flexible wires housed within a 3D-printed anterior shank module incorporated the Arduino Mini IMU. The Phidgets FlexiForce sensor was positioned under the heel and similarly routed to the central module, providing essential gait-event cues with minimal bulk and interference. The full set up of this prototype can be seen in Figure 18 as well as in Chapter 8.

Testing of this integrated design is documented in Chapter 8. The compression sock format enhanced wearer comfort and sensor stability, minimised previous issues such as fabric bunching and crosstalk, and aligned with the overarching principles of cost-effectiveness, portability, and clinical applicability for stroke populations.

6.6 Materials Selection and Specification

Table 4 summarises the materials and sensor options considered during device development, contrasting the initial selections with the final chosen components.

This table shows clear justification for each decision, highlighting the balance between signal fidelity, cost, portability, and clinical applicability that shaped the design pathway. The components described in the table are further explained through sections 6.6.1 to 6.6.5 below.

Table 4 - Illustrating the evolution of material and sensor selection across prototype phases. Final selections reflect alignment with device aims of low-cost accessibility, portability, and clinical relevance.

Component	Initial Option	Final Selection	Justification
Fabric base	Polyamide/elastane circular-knit (Footfalls & Heartbeats compression garment)	Same material integrated into compression sock design with copper-stitched electrodes	Provided Class 1 compression, durability under wash/wear cycles, and compatibility with textile-based, stitched, electrodes
Electrodes	Copper wire (<1 mm) stitched into fabric zones	Copper-alloy stitched electrodes aligned to SENIAM placement guidelines	Allowed integration of functional electrodes while maintaining comfort and mechanical stability
EMG sensor	Delsys Trigno wireless sensor (validated against Vicon Nexus)	SparkFun SEN-13723 MyoWare EMG board with Arduino interface	Reduced system cost while maintaining adequate signal fidelity; aligned with low-cost aim
IMU sensor	ActivPAL4 (validated against Vicon)	Arduino Nano 33 BLE Sense (Mini Plus) IMU	Supported low-cost, modular design; novel integration with fabric EMG + MyoWare board. To allow capture of SVA data.
Force sensor	Not included in Phase 1 testing	Phidgets FlexiForce Thin Force Sensor (100 kg capacity)	Enabled detection of heel strike and toe-off events; provided reliable temporal gait cues in a low-cost, unobtrusive format

6.6.1 Fabric base with copper stitched electrode

The fabric base was specified as a polyamide/elastane circular-knit material derived from the standard *Footfalls and Heartbeats* compression garment. When configured into a sock format, this material provides Class 1 graduated compression at the ankle (~18–32 mmHg), tapering proximally in accordance with established compression stocking guidelines.

This base fabric was selected for its combination of elasticity, durability, and clinical acceptability. In standard Class 1 compression applications, the material demonstrates durability targets of $\geq 80\%$ retention of nominal ankle pressure after 30 laundry cycles and $\leq 10\%$ drift in compression following two hours of light activity.

Copper wire electrodes (< 1 mm diameter) were stitched directly into the fabric, enabling the integration of a functional electrode array within the compression garment structure. This approach provided both secure skin contact and mechanical stability within the textile substrate.

The prototype 1 version of fabric EMG sensor can be seen in Figure 19 - Image of the Prototype 1 fabric base with copper stitched EMG electrode connected to a Delsys Trigno wireless unit as used in Stage 1 testing.” With this iteration of the design two fabric patches were required to cover the target SENIAM placement and allow data collection of the EMG signal passing through the target muscle.



Figure 19 - Image of the Prototype 1 fabric base with copper stitched EMG electrode connected to a Delsys Trigno wireless unit as used in Stage 1 testing.

6.6.2 Conductive Pathways and Electrode Zones

Electrodes were realised as knitted or embroidered conductive zones using fine copper-alloy wire (<1 mm). In the Stage 3 prototype, electrode positioning was aligned with SENIAM guidelines for surface EMG placement over the gastrocnemius lateralis, thereby ensuring comparability with standardised electrode placement protocols (Hermens et al., 2000).

Signal routing was achieved using a low-profile snap connector; in the first prototype this connector was positioned at the cuff, in the second and third prototypes the connector was approximately 1 inch anterior to the electrode location.

6.6.3 EMG sensor selection

During Phase 1 testing, the fabric electrode was validated against a Delsys Trigno wireless EMG sensor, with data collected via Vicon Nexus. This validation study, presented in Chapter 7, confirmed that the fabric electrodes were capable of capturing EMG signals comparable to those obtained with a gold-standard system. This set up can be seen in Figure 19.

Following this successful validation, and in line with the low-cost accessibility aim of the device (Table 3 - Mapping of Clinical Aims, Identified Gaps, and Device Requirements), the subsequent development stages replaced the Delsys system with a SparkFun SEN-13723 MyoWare EMG board with an Arduino platform.

This combination maintained adequate fidelity of EMG capture while significantly reducing system cost, thereby supporting scalability for clinical application.



Figure 20 - SparkFun SEN-13723 MyoWare EMG board

6.6.4 IMU sensor selection

In Phase 1, the ActivPAL4 IMU was employed alongside the fabric EMG electrode to allow monitoring of gait outcomes. As discussed in section 6.5.1, this device was selected due to previously demonstrated high reliability for detecting lower-limb inclination changes during postural transitions (Godfrey et al., 2015; Edwardson et al., 2017).

To further the low-cost design criteria, later development stages substituted the ActivPAL4 with an Arduino Nano 33 BLE Sense (Mini Plus) IMU. A scoping search found no peer-reviewed head-to-head validations of the Arduino Nano 33 BLE Sense IMU against Vicon for gait. However, several studies have validated comparable Arduino-compatible or low-cost microcontroller-based IMU systems incorporating similar inertial sensors (e.g., MPU-6050, MPU-9250, BNO055) against optical motion capture. For instance, Rodríguez-Martín et al. (2020) and Park and Yoon (2021) demonstrated that such systems can achieve strong agreement with laboratory-grade kinematics, supporting the feasibility of the present IMU-based approach. Nevertheless, the high degree of hardware and firmware variability inherent in Arduino-based platforms—one of their principal advantages in terms of flexibility and cost—also limits direct comparability across studies. As a result, no prior research has systematically examined the performance of a standardised Arduino Nano 33 BLE Sense configuration in conjunction with complementary sensing technologies. In this work, its integration with the MyoWare EMG board and fabric-embedded electrodes represents a novel and adaptable multimodal sensing system for capturing both kinematic and electrophysiological parameters during gait.

The Arduino platform was chosen for its affordability, modularity (up to 8 channels), and suitability for portable applications, thereby fulfilling the project's design objectives.

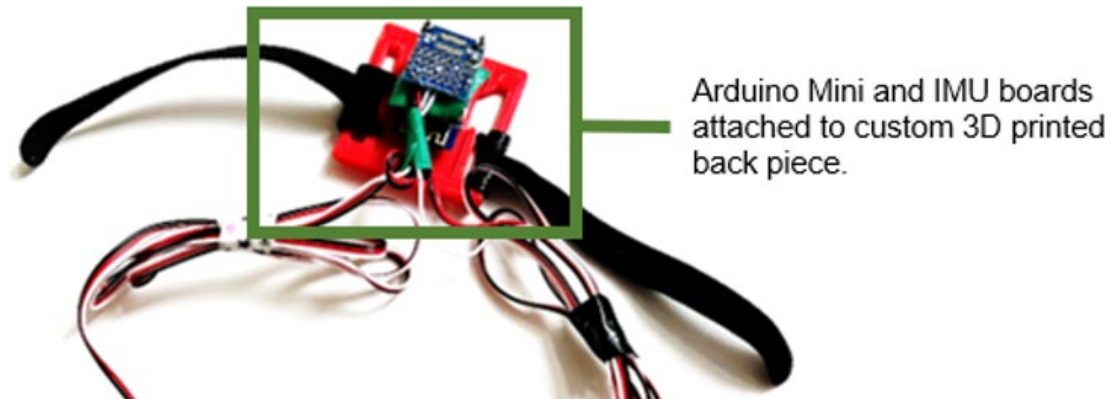


Figure 21 - Image showing Arduino Mini circuit board (IMU) connected to custom 3D printed base.

6.6.5 Additional Components

In order to complement the EMG and IMU components described in Sections 6.6.3 and 6.6.4, an 8 inch Phidgets FlexiForce 0-100lb Resistive Force Sensor (FSR) (as seen in Figure 22) was incorporated into the Stage 2 prototype to enable detection of discrete gait events, particularly heel strike and toe-off. The sensor consists of a thin, flexible polyester (Mylar) substrate with a silver conductive trace and resistive ink layer, producing a change in resistance proportional to applied load (Phidgets, 2017). Under no load, resistance is typically greater than 5 M Ω , decreasing to approximately 5 k Ω under maximum rated load, which allows it to output a responsive analogue signal suitable for digitisation and integration with Arduino-based hardware (Phidgets, 2017).



Figure 22 - Image of 8-inch Phidgets FlexiForce 0-100lb Resistive Force Sensor

The device's low profile, flexibility, and demonstrated durability of over one million load cycles at 50 lb impacts make it particularly suited for wearable applications where unobtrusive, repeatable event detection is required (Phidgets, 2017). While not intended for precise force quantification, the FlexiForce sensor provides reliable temporal cues, enabling low-cost approximation of force plate-derived gait event detection thresholds, such as the ~10 N criterion commonly used to define stance onset and termination (Winter, 2009).

However, such thin-film force sensors are not without limitations. Their performance is highly dependent on the stability of attachment, contact uniformity, and surface compliance, with signal drift and variability frequently arising from inconsistent sensor seating or material deformation (Al-Sayed et al., 2018; Prakash et al., 2019). Moreover, FlexiForce sensors are susceptible to damage or signal artefacts under shear loading, as lateral stresses can disrupt the conductive ink layers and alter sensitivity (Yazdan-Shahmorad & Sawan, 2016; Gouwanda & Senanayake, 2011).

These practical constraints necessitate careful mounting, calibration, and signal conditioning to ensure reproducibility. Nevertheless, when appropriately implemented, their integration strengthened the prototype's ability to capture clinically relevant gait events while maintaining the project's guiding principles of affordability, portability, and clinical relevance.

6.6.6 Safety and Contraindications

Compression levels were restricted to Class 1, consistent with UK compression garment guidelines, to ensure suitability for general use. This was done through collaboration with the manufacturer Footfalls Smartex Limited. (2025). Participant screening excluded individuals with contraindications such as self-reported severe peripheral arterial disease, acute deep vein thrombosis, or advanced neuropathy, as per clinical safety standards. Clinically, indications of these conditions were assessed for prior to initial device testing by the lead researcher, an HCPC registered Orthotist.

Throughout wear trials, skin compatibility of the conductive zones was monitored to detect potential irritation or discomfort. These safety considerations were embedded into the participant exclusion criteria and informed iterative refinements of the wearable prototype.

To satisfy the stated device requirement to be suitable for Stroke patients and therefore safe for use (6.1.1 Identified Clinical Needs and Gaps), during Phase 2 testing the Arduino IMU unit was housed in a 3D printed casing. This casing was designed to allow access to the functional switches and to insulate the user from any circuit board connections.

The casing served a secondary purpose to protect the hardware connections during use. An image of the Arduino IMU connected to 3D base section is shown in Figure 21 and an image of the full housing can be seen in Figure 23.

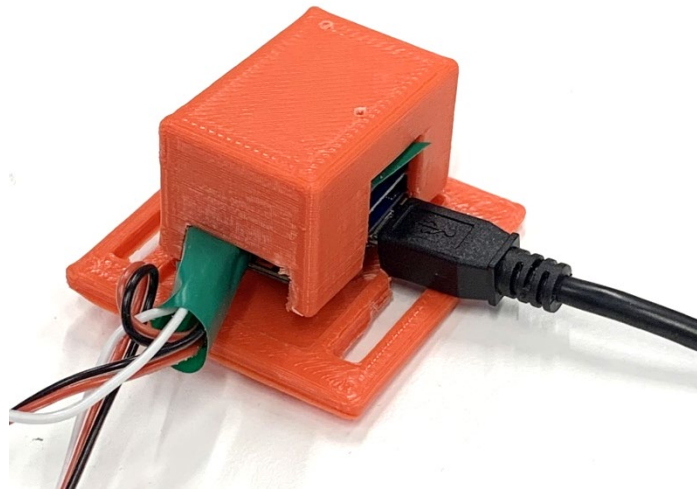


Figure 23 - Image showing the full 3D printed casing for the Arduino Mini IMU unit.

6.7 Summary of Prototype Development Process

The prototype development process followed a structured, iterative pathway that was firmly anchored in the clinical needs and device aims identified in Chapters 2 and 5. Each design decision was made with direct reference to the five overarching aims, ensuring a strong and traceable link between identified clinical priorities and technical implementation.

First, the aim of low-cost and accessibility was addressed by transitioning away from reliance on high-cost commercial systems such as the Delsys Trigno and Vicon Nexus, used only in early proof-of-concept testing, towards the integration of MyoWare EMG boards and an Arduino Pro Mini IMU.

This combination provided reliable data capture at a fraction of the cost of laboratory-based solutions, thereby fulfilling the requirement for a device that

could be realistically deployed in community and hospital rehabilitation settings.

Second, the requirement for portability and integration guided the consolidation of hardware into a single wearable system. The development moved from separate fabric patches with external cabling in Stage 1 to an integrated compression garment housing stitched electrodes, connected via low-profile snap connectors to lightweight processing modules. The final prototype combined EMG, IMU, and force sensing within a compact form, reducing set-up time and enabling smooth incorporation into routine clinical workflows.

Third, to ensure the capture of clinically relevant outputs, the system was designed to provide gait speed, cadence, and shank-to-vertical angle (SVA), reflecting both the literature and clinician-identified priorities (Perry et al., 1995; Awad et al., 2016; Owen, 2004; Kerr et al., 2020; Tudor-Locke et al., 2018). The addition of a Phidgets FlexiForce sensor in Stage 2 further strengthened the system's ability to detect discrete gait events such as heel strike and toe-off, facilitating reliable segmentation of gait cycles and enabling more precise output measures.

Fourth, the need for objective orthotic assessment was incorporated by combining EMG with kinematic and force data. This provided clinicians with quantitative information on muscle activation patterns, joint kinematics, and temporal gait events, creating an evidence base to support more accurate ankle-foot orthosis (AFO) alignment and potentially informing person-centred timing of intervention (Mulroy et al., 2010; Burpee & Lewek, 2015).

Finally, the principle of suitability for stroke populations shaped the evolution of the garment design. Stage 1 highlighted discomfort and instability from separate patches, and Stage 2 identified issues of fabric bunching and

electrode crosstalk. These issues were resolved in Stage 3 through the adoption of a Grade 1 compression sock base, which provided consistent electrode placement, reduced motion artefact, and ensured safety and comfort. This addressed concerns regarding usability in individuals with sensory deficits or circulatory comorbidities, maintaining alignment with patient safety standards.

Figure 24 illustrates the progression of prototype development mapped against the identified clinical aims. Stage 1 addressed only a subset of requirements, with limited achievement of clinically relevant outputs. Stage 2 demonstrated partial fulfilment of all aims through integration of EMG, IMU, and force sensing, though fabric usability limitations persisted. Stage 3 achieved full alignment with all five aims.

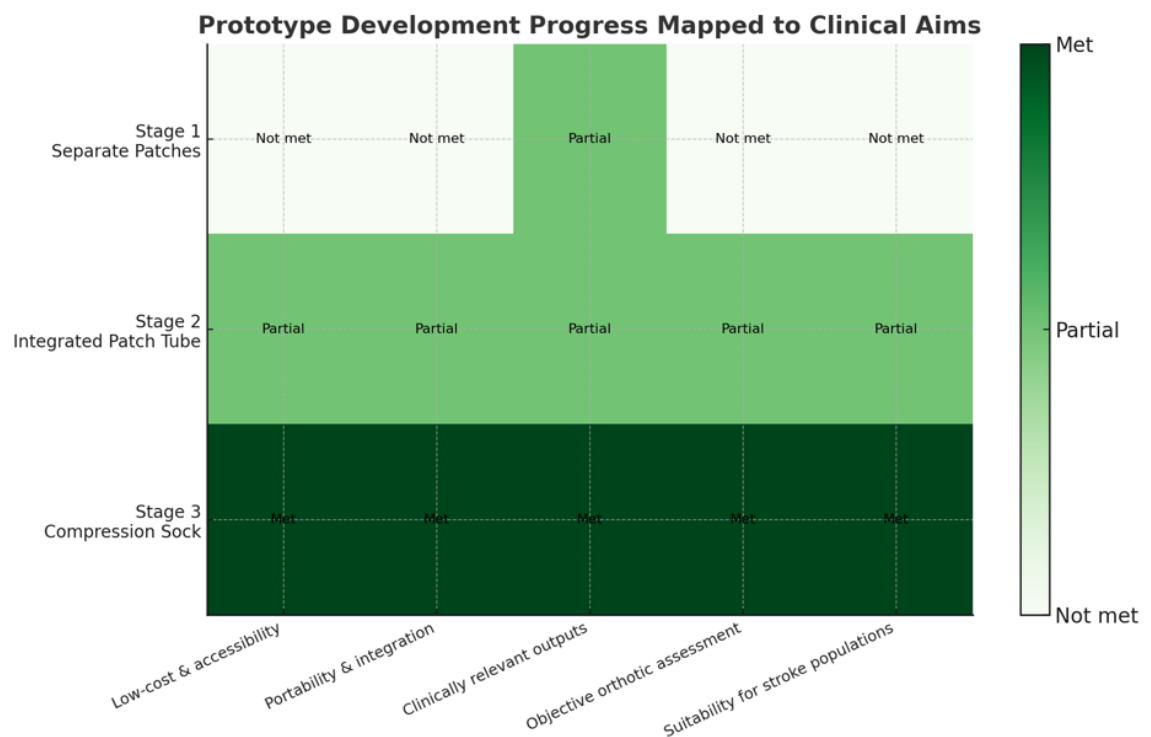


Figure 24 - Prototype development process mapped against identified clinical aims. Shading indicates whether aims were unmet, partially met, or fully achieved at each stage.

In summary, the development of the prototype demonstrates a clear progression through design iterations in line with the identified design protocols. As per Human-Centred Evaluation practices, each stage responded to the feedback, identified clinical gaps and device aims, ensuring that the final system directly addressed barriers to current practice and aligned with the practical realities of stroke rehabilitation.

The following chapter describes in full the research study that utilised Prototype 1 and examines the clinical outcomes in relation to the design parameters discussed here, thereby establishing the evidence base that informed later iterations of the device.

Chapter 7 – Proof-of-Concept Cohort Case Study to validate the Novel Fabric EMG Sensor and activPAL Activity Monitor.

This chapter details the first participant study undertaken which was a proof-of-concept, cohort case study which aimed to validate a novel fabric EMG sensor design against the industry gold standard. In line with the thesis aims, this study sought to investigate tools which could be used to improve the functional and descriptive analysis of gait overall and of Stroke specifically.

Acknowledgements – I would like to thank BME Masters students Miss Concannon and Miss L'Ange for their assistance in data collection and analysis during this study.

7.1 Introduction

According to the National Institute for Health and Care Excellence (NICE, 2019a) and the Stroke Association's 2018 *State of the Nation* report, more than 100,000 people in the UK experience a stroke each year, resulting in approximately 38,000 deaths. Stroke therefore remains the fourth leading cause of death nationally. Mortality rates have declined substantially, by nearly 50% between 1990 and 2010, largely due to advances in acute management (Stinear et al, 2017).

However, survival has created a new challenge: there are now more than 1.2 million stroke survivors in the UK, with a growing proportion living with long-term disability and an associated societal cost of £26 billion annually (NICE, 2019a).

Functional disability is common, with almost 65% of survivors discharged from hospital experiencing lasting impairments such as hemiparesis, visual field loss, dysphasia, or sensory deficits. Of these, hemiparesis is among the most disabling sequelae, affecting up to 65% of the stroke population (Cauraugh & Kim, 2003). When the lower limb is involved, reduced voluntary muscle control compromises walking ability and restricts independent mobility. Restoration of gait is therefore a central rehabilitation goal, with improvements in both walking pattern and velocity consistently prioritised as key outcomes (Alingh, Groen, Van Asseldonk, Geurts, & Weerdesteyn, 2020).

As highlighted in Chapter 2, hemiparetic gait is typically inefficient, marked by abnormal synergies, reduced propulsion from the paretic limb, and reliance on compensatory strategies such as hip hiking or circumduction. While these strategies enable forward progression, they come at the cost of elevated energy expenditure and reduced endurance (Awad et al., 2015). Such inefficiencies not only limit community participation but also contribute to increased risk of falls and recurrent stroke due to heightened fatigue. Accurate measurement of gait mechanics is therefore critical to rehabilitation research and clinical practice.

Orthotic intervention, most notably ankle–foot orthoses (AFOs), remains a widely prescribed strategy to manage swing-phase deficits such as foot drop, thereby reducing tripping risk and improving safety (Mulroy et al., 2010). Yet debate persists over whether long-term AFO use may inadvertently constrain neuroplastic recovery potential in some individuals. This aligns with findings from the Chapter 5 questionnaire study, which indicated considerable variation in clinical practice across the UK, including inconsistent use of gait laboratories and divergent attitudes toward orthotic prescription. These discrepancies reinforce the importance of developing more reliable and accessible gait assessment tools to guide treatment.

Historically, gait speed has been used as the primary rehabilitation outcome measure. While self-selected gait speed provides a simple and widely accepted metric, it fails to account for compensatory mechanisms that can artificially inflate performance without reflecting underlying neuromuscular recovery (Awad et al., 2016; Burpee & Lewek, 2015; Jonkers et al., 2009; Kerr et al., 2020; D. C. Kerrigan et al., 1998; Nadeau et al., 1999). Furthermore, as highlighted by Langhorne et al. (2018), inconsistencies in the reporting and interpretation of gait speed data hinder cross-study comparison and limit its predictive value. These issues were also observed in the literature synthesis presented in Chapter 2, where heterogeneous outcome measures restricted direct comparability across rehabilitation trials.

This context underscores the rationale for the present study. As outlined in the overarching aims of this thesis (Chapter 3), there is a pressing need for new tools that enable detailed, real-world analysis of gait mechanics and muscle activation in stroke survivors. By integrating the findings from the device development work in Chapter 6 with empirical validation in this proof of concept study, the thesis advances toward a practical solution: wearable systems capable of capturing meaningful biomechanical and EMG data in both laboratory and clinical environments.

7.3 Aim

Gold-standard approaches such as 3D motion capture and surface EMG provide detailed insights but are resource-intensive and largely unavailable in day-to-day practice (Burpee & Lewek, 2015; Langhorne et al., 2018). At the same time, clinicians are increasingly expected to make decisions about orthotic prescription and gait training with limited access to objective biomechanical data.

Wearable technologies have the potential to close this gap, provided their outputs can be demonstrated to match the accuracy of laboratory-based systems. Textile EMG sensors and small inertial monitors are particularly promising, offering the possibility of in-brace monitoring and continuous data collection beyond specialist laboratories.

This study therefore set out two aims:

- Primary aim – To pilot-validate the Footfalls fabric EMG sensor against the Delsys Trigno gold-standard. This will determine whether a textile-based electrode can accurately capture lower-limb muscle activity, including when worn beneath an AFO.
- Secondary aim – To examine the accuracy of shank-to-vertical angle (SVA) derived from the activPAL monitor compared with Vicon motion capture data. An error threshold of $<5^{\circ}$ was set as the criterion for acceptability.

By addressing these aims, the study builds on the device development work presented in Chapter 6 and contributes to the overall thesis objective of validating practical, real-world tools for post-stroke gait assessment.

7.4 Method

7.4.1 Study Design

This study was conducted as a proof-of-concept cohort case study using healthy, non-stroke-affected participants. The purpose was to generate preliminary evidence on the performance of a fabric-based EMG sensor (Footfalls, 2020) and the activPAL activity monitor (PAL Technologies, 2020),

benchmarked against gold-standard Vicon motion capture and Delsys Trigno EMG. The staged design enabled testing under controlled laboratory conditions prior to translation to stroke populations. Accordingly, the study is framed as a proof-of-concept investigation, with the small sample size ($n = 4$) intentionally used to demonstrate system functionality, signal integrity, and protocol implementation, rather than to support statistical inference or generalisation.

7.4.1.2 Ethics

Ethical approval for this study was obtained from the University of Strathclyde Ethics Committee in accordance with institutional research governance procedures (University of Strathclyde Ethics Reference: UEC21/54 Kerr/L'Ange/Campbell – see Appendix C.2). All participants received written information about the study and provided informed consent prior to participation – Appendix C.3.

7.4.2 Participants

Healthy adult volunteers were recruited from the postgraduate and staff community at the University of Strathclyde. Participants were required to meet basic eligibility criteria to ensure safe participation and to minimise confounding factors during gait analysis. All provided informed consent prior to involvement in accordance with university policies and Good Practice Guidelines.

7.4.2.1 Inclusion criteria

- Age ≥ 18 years.
- Independently mobile with unimpaired gait.

- Ability to follow simple instructions.
- Ability to communicate in English.
- Capacity to provide informed consent.

7.4.2.2 Exclusion criteria

- Neurological or musculoskeletal conditions affecting walking ability.
- Cardiovascular conditions or other health issues restricting participation in mild to moderate exercise (e.g., cardiac conditions exacerbated by exertion).
- Current use of medications that may affect walking ability or comprehension.
- Uncorrected vision or hearing impairments.
- History of vestibular or balance disorders.
- Dermatological conditions aggravated by hypoallergenic tape.
- Pregnancy.
- Current or recent COVID-19 symptoms, active diagnosis, or close contact exposure.

7.4.3 Protocol

Data collection was undertaken in the temporary CAREN gait laboratory, located within the National Centre for Prosthetics and Orthotics (Curran Building) at the University of Strathclyde. A 12-camera Bonita Vicon motion capture system (Vicon Motion Systems Ltd., Oxford, UK) was used to capture kinematic, kinetic, and spatiotemporal gait data, synchronised with a split-belt force-instrumented treadmill (Motek Medical, Amsterdam, Netherlands). Marker trajectories and force data were captured through Vicon Nexus software for subsequent analysis.

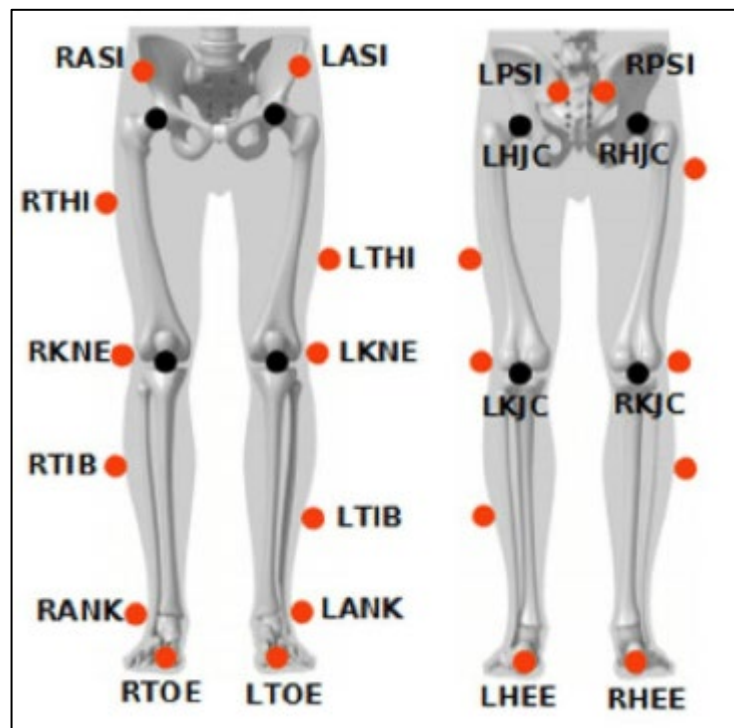


Figure 25 - Diagram to illustrate placement of reflective markers on trial participant (Baudet et al., 2014)

Retroreflective markers were applied using the Plug-in-Gait lower-limb model (Baudet et al., 2014) and secured with hypoallergenic tape (Figure 25). The Vicon system recorded at 100 Hz, while the force-instrumented treadmill sampled at 2000 Hz.

As seen in Figure 26 below, to promote a natural gait pattern, the D-Flow software (Motek Medical, Amsterdam, Netherlands) projected a simple virtual pathway onto front-facing screens.

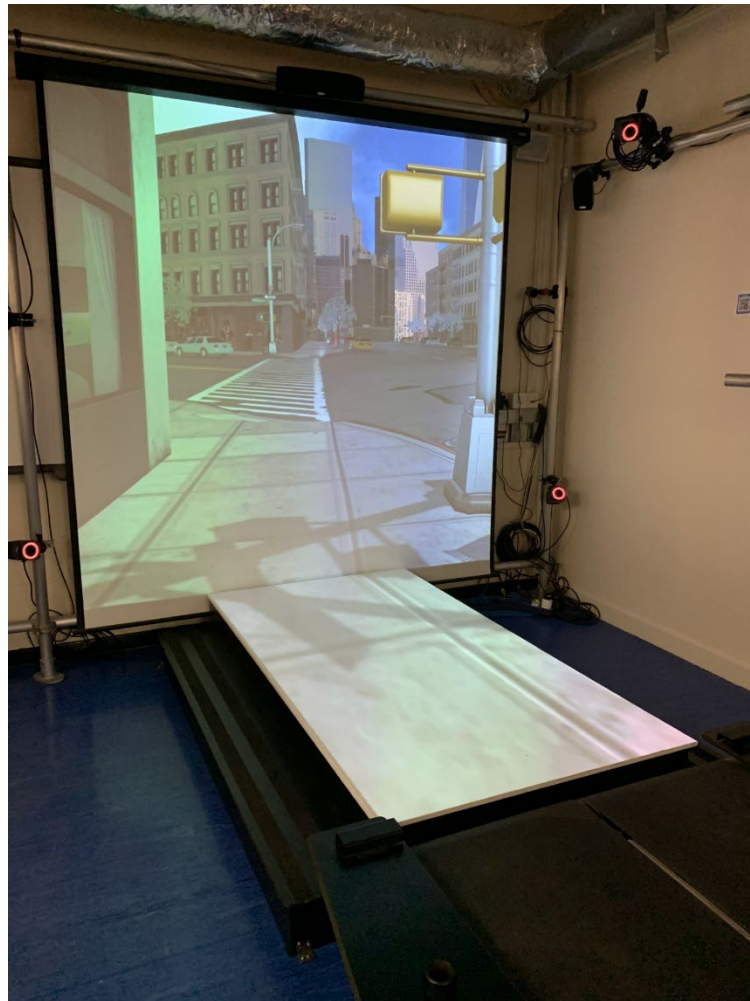


Figure 26 - Image of the D-Flow software (Motek Medical, Amsterdam, Netherlands. N.d.) projecting a virtual pathway onto a front-facing screen.

All treadmill trials were performed with participants wearing a safety harness, consistent with the approved risk assessment reviewed by the University Ethics Committee, to minimise risk of falls.

Participants were asked to walk at three self-selected speeds—slow, normal, and fast—to capture the variability and adaptability of gait performance across functional ranges. This approach is consistent with the evidence discussed in

Chapter 2, which highlights walking speed as both a sensitive marker of recovery and a widely used clinical outcome measure (Bohannon, 1997; Perry et al., 1995). Post-stroke gait is often characterised by reduced self-selected speed, limited capacity to modulate velocity, and impaired propulsion mechanics (Section 2.3.1).

By including slow, normal, and fast conditions, the study accounted for these deficits while also exploring whether neuromuscular and kinematic markers (e.g., propulsion, shank-to-vertical angle, gastrocnemius activation) respond differently across functional speed ranges.

Furthermore, stroke survivors frequently use compensatory strategies at slower speeds, whereas higher speeds may demand greater engagement of plantarflexors and improved coordination (Section 2.3.3.1). Evaluating across speeds therefore allowed differentiation between compensatory performance and genuine recovery, a distinction emphasised as critical in recent literature (Kwakkel et al., 2019; Balasubramanian et al., 2007).

7.4.3.1 Testing Stage 1

During the initial testing stage, surface EMG was collected from the lateral head of the left gastrocnemius muscle using a Delsys Trigno wireless EMG sensor (Delsys, Boston, USA). Placement was guided by SENIAM recommendations (Stegeman & Hermens, 2007) (Figure 26).

As previously discussed in section 6.3.2.3: the lateral gastrocnemius was selected for several reasons. Functionally, the gastrocnemius is a biarticular plantarflexor that contributes critically to forward propulsion during terminal stance and push-off, roles that are often impaired in post-stroke gait (Bowden et al., 2006; Awad et al., 2016; Neptune, Kautz, & Zajac, 2001). Anatomically, the lateral head provides a broad, superficial belly that facilitates reliable

electrode placement and reduces crosstalk compared to the medial head or soleus.

Placement on the lateral head also avoided the popliteal fossa, thereby minimising motion artefacts and improving participant comfort. In the context of orthotic intervention, recording from the gastrocnemius provides a meaningful indicator of how ankle–foot orthoses and shank-to-vertical angle alignment influence plantarflexor activation and propulsion mechanics (Bregman et al., 2012; Kerr et al., 2020). Finally, the use of this site aligns with prior EMG studies in stroke gait rehabilitation (Clark et al., 2010; Cheung et al., 2012), ensuring comparability of findings with the wider literature.

According to the guidelines, surface EMG electrodes for the lateral head of the gastrocnemius should be positioned at one-third of the distance along the line connecting the head of the fibula to the heel (Figure 27). The electrodes should be aligned in the direction of this line to optimise signal capture from the target muscle fibres and reduce crosstalk from adjacent muscles (SENIAM, n.d.).

Two ActivPAL IMU devices were attached to the participant using self-adhesive tape as with the Delsys electrode. One was positioned on the anterior shin and the second on the anterior thigh. Each activPAL4 contained an accelerometer and magnetometer sampling at 20 Hz, with dimensions of 5.3 × 3.5 × 0.7 cm and a mass of 15 g.

Participants completed treadmill walking at three speeds: self-selected comfortable, slow, and fast.

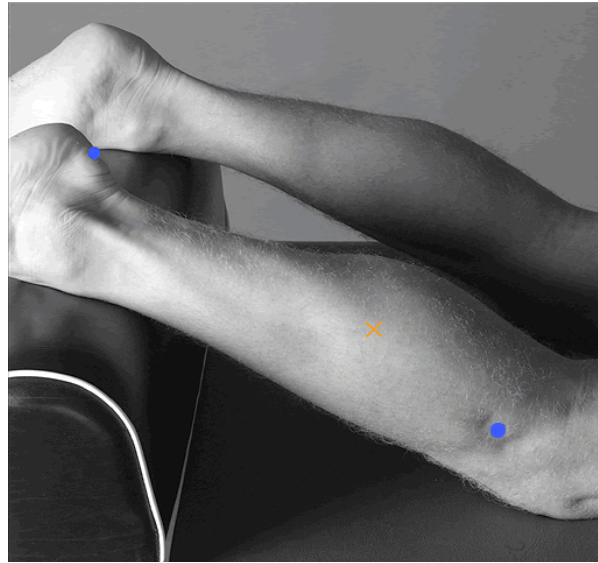


Figure 27 - Electrode placement for the lateral gastrocnemius according to SENIAM guidelines. SENIAM. (n.d.)

7.4.3.2 Testing Stage 2

In the second testing stage, the Delsys wireless sensor was replaced with a novel prototype fabric EMG sensor developed by Footfalls & Heartbeats (Nottingham, UK). The device consisted of a knitted textile with integrated conductive fibres (as described in Chapter 6). A Delsys snap-lead connector enabled data capture via Vicon Nexus. The textile electrode was secured to the participant with an elastic strap and hypoallergenic tape.

During this testing the ActivPAL IMUs remained in place.

Walking trials were repeated at slow, comfortable, and fast speeds to allow direct comparison across sensors.

7.4.3.3 Testing Stage 3



Figure 28 - An image of a stock, 3/4 length footplate Ankle Foot Orthosis (AFO) from Chaneco, (2021).

For the final testing stage, participants were fitted with a posterior leaf spring ankle–foot orthosis (PLS AFO; Chaneco, 2021), constructed from polypropylene with a $\frac{3}{4}$ length footplate (Figure 28).

In contemporary stroke rehabilitation, the clinical gold standard for ankle–foot orthosis provision is typically a custom-made solid ankle AFO, individually tuned to optimise both stance and swing phase gait mechanics. These devices are commonly aligned and adjusted to improve sagittal plane efficiency, facilitate foot clearance, enhance knee stability, and influence overall propulsion during walking (Mulroy et al., 2010; Bregman et al., 2012; Adiputra et al., 2019). However, custom orthotic intervention requires multiple clinical appointments, casting or digital scanning procedures, and individual manufacturing processes.

Off-the-shelf (OTS) orthoses represent a commonly used alternative where patients present with less complex biomechanical requirements. Such devices allow faster and lower-cost intervention, particularly for individuals with more isolated swing phase deficits who do not demonstrate significant coronal plane ankle instability, substantial knee hyperextension, or complex comorbidities requiring bespoke orthotic management. Although OTS devices are generally less durable and less adjustable than custom orthoses, they can still provide clinically appropriate support for selected users and are frequently used within early rehabilitation pathways and interim orthotic management.

The OTS PLS AFO selected for this study was therefore chosen on pragmatic and methodological grounds. The time, cost, and infection-control implications associated with producing custom orthoses for all participants during the COVID-19 restrictions made bespoke fabrication impractical and introduced unnecessary additional clinical contact. The selected device was considered sufficient to reduce natural sagittal plane deviation in non-affected participants, thereby approximating some of the biomechanical influence that a tuned custom AFO would exert on ankle mechanics in a stroke population.

The inclusion of an ankle–foot orthosis during testing was also justified on both clinical and methodological grounds. As highlighted in Chapter 2, AFOs remain among the most widely prescribed orthotic interventions for stroke survivors, with evidence demonstrating improvements in foot clearance, energy efficiency, and gait stability (Mulroy et al., 2010; Bregman et al., 2012; Adiputra et al., 2019). Their routine use within rehabilitation practice means that any novel gait assessment or monitoring technology intended for clinical translation must demonstrate compatibility with orthotic intervention. Excluding AFO use would therefore reduce the ecological validity of the study and fail to reflect the conditions under which many stroke survivors ambulate.

From a biomechanical perspective, AFOs directly influence propulsion, knee kinematics, and ankle joint mechanics (Harper et al., 2014; Bleyenheuft et al., 2013). Incorporating orthotic use within the test protocol therefore enabled assessment of whether the novel device could reliably capture neuromechanical signals in the presence of orthotic tuning effects, particularly gastrocnemius EMG activity. Finally, the trimlines and material properties of the selected OTS device allowed fitting within participants' own standard sports footwear, avoiding the need for custom footwear and supporting a more practical representation of real-world orthotic use.

Note - During the data collection for Testing stage 3 the ActivPAL IMUs were removed from the patient due to interference with the AFO placement. IMU data was not collected for this cohort as the focus shifted to testing practicality and useability of the novel EMG device in situ with an AFO.

The Delsys EMG sensor remained in place connected to the fabric electrodes as in Testing stage 2 (section 7.4.4.2).

7.4.4 Data Analysis

Data were processed and analysed using Vicon Nexus (version 2.10.1; Vicon Motion Systems Ltd., Oxford, UK), MATLAB R2020a, and MATLAB (MathWorks, 2021). Software licences for all software were provided through the University of Strathclyde.

7.4.4.1 Shank-to-Vertical Angle (SVA)

SVA was calculated for each participant at slow, comfortable, and fast walking speeds without the AFO. For Vicon, SVA was computed from reflective marker trajectories based on the Plug-in-Gait model, with angle calculated relative to

the vertical axis in the sagittal plane. For activPAL4, SVA values were provided directly by PAL Technologies (Glasgow, UK) through proprietary processing routines. Although the underlying algorithms were not accessible, activPAL devices are known to estimate segment orientation using triaxial accelerometer and magnetometer signals.

For example: The SVA can be estimated directly from the accelerometer channels of the activPAL without applying an explicit rotation matrix. Assuming the sensor's z-axis is aligned vertically and the x-axis aligned along the anterior–posterior axis of the shank, the pitch of the shank relative to vertical can be expressed as:

$$SVA(t) = \arctan \left(\frac{a_x(t)}{a_z(t)} \right)$$

Where $a_x(t)$ is the anterior–posterior acceleration and $a_z(t)$ is the vertical acceleration at time t .

To normalise across participants and sensor placements, the standing baseline can be subtracted:

$$SVA_{adj}(t) = SVA(t) - SVA_{standing}$$

Where $SVA_{standing}$ is the mean angle recorded during a quiet standing calibration.

Stride-level summaries can then be derived as:

$$SVA_{stance} = \frac{1}{N} \sum_{i=1}^N SVA_{adj}(i)$$

With N representing the number of samples during stance.

Accelerometer data provide static orientation relative to gravity, while magnetometer input enables correction for drift and alignment with the vertical axis. In practice, the accelerometer detects the direction of the gravitational vector, which allows estimation of sensor tilt when the segment is stationary or moving slowly. Because accelerometers also capture dynamic accelerations during motion, their output alone is prone to noise and drift. The magnetometer complements this by measuring the Earth's magnetic field, providing a consistent heading reference that stabilises orientation estimates over time.

Through sensor fusion algorithms data from both sensors are integrated to continuously align the device's coordinate system with the global vertical and horizontal axes. These combined signals allow accurate reconstruction of limb segment orientation, from which the shank-to-vertical angle (SVA) is derived. Mean error values for the activPAL4 relative to Vicon were then calculated for each participant and condition to quantify measurement accuracy.

Additionally, agreement between activPAL4 and Vicon measurements of shank-to-vertical angle (SVA) was assessed using Bland–Altman analysis. This method is widely employed in validation studies to quantify the level of agreement between two measurement systems by plotting the mean difference (bias) alongside the 95% limits of agreement. For each walking condition (comfortable, slow, and fast), participant-level differences were

plotted against the mean of the two systems to provide a descriptive assessment of measurement agreement.

7.4.4.2 Electromyography (EMG)

Raw EMG data were pre-processed in MATLAB. A sixth-order Butterworth band-pass filter (20–400 Hz) was applied to reduce low-frequency motion artefact and high-frequency noise. To suppress electrical interference, a 60 Hz notch filter was applied. The signals were then full-wave rectified and smoothed using a moving average window of 200 frames, producing an EMG envelope suitable for identifying muscle activation patterns (De Luca, 1997).

From these processed signals, peak gastrocnemius activation timing (as a percentage of gait cycle) and amplitude were extracted for comparison between the Delsys Trigno and Footfalls fabric sensors across conditions.

EMG data were processed to extract both temporal and amplitude-based characteristics of gastrocnemius activation. Peak activation timing was expressed as a percentage of the gait cycle for each condition (Trigno, fabric sensor, fabric sensor with AFO). Where multiple activation bursts occurred within a cycle, both primary and secondary peaks were identified and recorded. In addition to peak timing, onset and offset of muscle activity were determined using threshold-based detection methods to provide activation windows across the gait cycle.

To enable cross-condition comparison, the variability of peak timings across cycles was quantified using standard deviation and coefficient of variation. All analyses were conducted descriptively, in line with the feasibility design and small sample size ($n=4$). The primary objective was to characterise consistency and agreement between devices rather than to test statistical hypotheses.

7.4.4.3 Data Handling and Management

All raw and processed data were stored on secure, password-protected University of Strathclyde servers in accordance with the University's Research Data Management Policy. Handling, storage, and retention procedures followed the requirements outlined in the approved ethics application for this study and complied with the General Data Protection Regulation (GDPR, 2018).

As described in Chapter 5, participant data were anonymised at the point of collection, with unique identifiers replacing personal information. Data were used exclusively for the purposes of this research and will be retained in line with institutional guidance.

7.4.5 Aims – Method Alignment Table

The alignment of aims and methods in Chapter 7 demonstrates how the proof-of-concept study directly addressed the overarching objectives of this thesis. The primary aim focused on validating the novel fabric EMG sensor against gold-standard systems, thereby providing initial evidence of technical performance and feasibility for in-brace monitoring. The secondary aim evaluated the activPAL monitor as a pragmatic alternative to laboratory kinematic analysis, assessing its accuracy in estimating shank-to-vertical angle relative to Vicon.

Together, these aims ensured that the methodological approach remained consistent with the developmental pathway outlined in Chapter 3, linking device design (Chapter 6) with initial feasibility testing. Table 5 below illustrates this achievement.

Table 5 - Depicting the Aims - Method alignment and mapping to Thesis Aims from Chapter 3.

Aim	Methodological Alignment	Link to Thesis Objectives
Primary Aim – To pilot-validate the Footfalls fabric EMG sensor against the industry gold-standard Delsys Trigno EMG system, using Vicon motion analysis as the reference.	- EMG collected from the lateral gastrocnemius using both Delsys Trigno and Footfalls fabric sensor across three walking speeds (slow, comfortable, fast). - Standardised electrode placement following SENIAM guidelines. - Data synchronised with Vicon Nexus (2.10.1). - EMG pre-processed in MATLAB using band-pass (20–400 Hz), 60 Hz notch filter, rectification, and moving average smoothing. - Peak timing and amplitude of gastrocnemius activation compared across systems, including condition with posterior leaf spring AFO.	Contributes to the overall thesis aim of developing and validating novel wearable technologies for post-stroke gait assessment (Objective 2). Provides preliminary validation of fabric-based EMG suitable for use with orthoses, directly supporting the device design and feasibility testing pathway.
Secondary Aim – To pilot-validate shank-to-vertical angle (SVA) estimates derived from the activPAL wearable monitor against Vicon-derived kinematic data, with <5° error defined as acceptable.	- Two activPAL4 monitors placed on anterior thigh and tibia during trials without AFO. - Vicon marker trajectories used to calculate gold-standard SVA. - ActivPAL SVA outputs obtained from proprietary software provided by PAL Technologies. - Mean error between activPAL and Vicon SVA calculated for each participant at each walking speed.	Aligns with the thesis objective of identifying practical, real-world tools for gait assessment (Objective 3). Demonstrates the potential of low-cost, portable IMUs for estimating gait parameters that are currently restricted to laboratory-based analysis.

In summary, the methodological approach adopted in this proof-of-concept study ensured that both the novel fabric EMG sensor and the activPAL monitor were evaluated against gold-standard laboratory systems under controlled conditions. By addressing the defined aims through systematic testing across multiple walking speeds and device configurations, the study generated preliminary evidence of technical validity and practical feasibility.

The following section presents the results of these analyses, focusing on the accuracy of activPAL-derived shank-to-vertical angle and the comparability of fabric EMG outputs with conventional electrodes.

7.5 Results

Four participants (2 female, 2 males, mean age ($\pm$ SD) = 28.5 ($\pm$ 5.9) years) were recruited from within the Biomedical Engineering Department in the University of Strathclyde. No issues occurred during the data collection, and no ill-effects were reported by the participants.

7.5.1 Peak Shank to Vertical angle (SVA) (activPAL and Vicon comparison)

7.5.1.1 *Comfortable walking speed*

At the comfortable pace, the mean peak SVA recorded by Vicon was 56.96° ($\pm 1.94^\circ$), compared with 52.80° ($\pm 8.09^\circ$) from the activPAL4. This produced a mean error of 4.16° ($\pm 9.27^\circ$). However, individual participant data showed considerable variability. Participant 1 exhibited the largest discrepancy, with activPAL underestimating SVA by 13.89° (Vicon: $59.08^\circ \pm 0.65$; activPAL: $45.19^\circ \pm 3.03$). Participant 2 showed the opposite trend, with activPAL overestimating by 7.50° (Vicon: $55.10^\circ \pm 1.08$; activPAL: $62.60^\circ \pm 7.48$).

By contrast, Participants 3 and 4 demonstrated smaller errors of 1.74° and 8.76° respectively. Overall, activPAL accuracy at comfortable speed varied widely across individuals.

7.5.1.2 Slow walking speed

During slow walking, mean peak SVA values were 54.44° ($\pm 2.08^\circ$) for Vicon and 39.09° ($\pm 4.67^\circ$) for activPAL, resulting in the largest mean error of 19.30° ($\pm 5.42^\circ$). All four participants showed substantial underestimation by activPAL. The largest discrepancies were observed in Participant 1 (22.07° error) and Participant 4 (23.36° error), while Participants 2 and 3 recorded errors of 17.70° and 14.06° respectively.

These results indicate that activPAL performance was least reliable at slower walking speeds, where systematic underestimation of SVA was observed across the cohort.

7.5.1.3 Fast walking speed

At the fast pace, mean peak SVA values were 59.89° ($\pm 1.42^\circ$) for Vicon and 61.39° ($\pm 7.17^\circ$) for activPAL. In contrast to the other conditions, activPAL slightly overestimated the group mean, producing a small negative mean error of -1.49° ($\pm 5.80^\circ$). Participant-level analysis revealed mixed patterns: Participant 1 showed an underestimation of 4.89° , while Participant 2 demonstrated the largest overestimation of 9.00° .

Participants 3 and 4 recorded near-perfect agreement, with errors of 0.38° and -2.23° respectively. These findings suggest that activPAL accuracy was most consistent at higher walking speeds.

A full breakdown of mean peak SVA values and associated errors for each participant and walking speed is presented in Table 6.

		Vicon Mean Peak SVA ($\pm$SD)	ActivPAL Mean Peak SVA ($\pm$SD)	Mean error
Comfortable walking speed	Participant 1	59.08° ($\pm$ 0.65)	45.19° ($\pm$ 3.03)	13.89°
	Participant 2	55.10° ($\pm$ 1.08)	62.60° ($\pm$ 7.48)	-7.50°
	Participant 3	55.50° ($\pm$ 0.50)	53.76° ($\pm$ 2.13)	1.74°
	Participant 4	58.46° ($\pm$ 0.51)	49.66° ($\pm$ 5.42)	8.76°
	Average	56.96° ($\pm$ 1.94)	52.80° ($\pm$ 8.09)	4.16° ($\pm$ 9.27)
Slow walking speed	Participant 1	56.06° ($\pm$ 0.57)	36.31° ($\pm$ 2.35)	22.07
	Participant 2	51.84° ($\pm$ 1.40)	38.74° ($\pm$ 2.90)	17.70°
	Participant 3	54.17° ($\pm$ 0.68)	45.43° ($\pm$ 2.53)	14.06°
	Participant 4	55.69° ($\pm$ 0.40)	35.87° ($\pm$ 2.99)	23.36°
	Average	54.44° ($\pm$ 2.08)	39.09° ($\pm$ 4.67)	19.30° ($\pm$ 5.42)
Fast walking speed	Participant 1	61.73° ($\pm$ 0.26)	56.84° ($\pm$ 4.59)	4.89°
	Participant 2	58.92° ($\pm$ 0.77)	67.92° ($\pm$ 9.09)	- 9.0°
	Participant 3	58.59° ($\pm$ 0.41)	58.21° ($\pm$ 4.16)	0.38°
	Participant 4	60.53° ($\pm$ 0.32)	62.76° ($\pm$ 3.69)	-2.23°
	Average	59.87° ($\pm$ 1.42)	61.39° ($\pm$ 7.17)	- 1.49° ($\pm$ 5.80)

Table 6 - Comparison of peak mean SVA angles recorded in Vicon and with the ActivPAL sensor for all four participants walking at their comfortable, slow and fast speeds.

7.5.2 EMG

7.5.2.1 Phase 1 – Delsys Trigno EMG sensor

During the initial phase with the wireless Trigno sensor, Participant 1 recorded a single peak gastrocnemius activation at 48.9% of the gait cycle. Participant 2 showed a primary peak at 34%, with additional late-cycle activations at 85% and 95%. Participant 3 recorded an early peak at 25% and a secondary peak around 90%. Participant 4 demonstrated a late-cycle peak at 60%, accompanied by unusually high activity during the first half of the gait cycle, with amplitudes up to 4.6×10^{-3} V (approximately three times higher than expected baseline values).

7.5.2.2 Phase 2 – Fabric EMG sensor with snap-lead connector

Usable Vicon data were obtained for Participants 1 and 2. A subsequent hardware communication fault prevented further recordings, and replication was not feasible within the study timeframe. Participant 1 demonstrated peak activation at 41.6% of the gait cycle, while Participant 2 recorded a peak at 40%. These results were consistent with those obtained using the Trigno sensor and align with established normative values for gastrocnemius activation.

7.5.2.3 Phase 3 – Fabric EMG sensor worn in combination with AFO

When the fabric sensor was worn under an AFO, Participant 1 demonstrated peak activation at 45% of the gait cycle and an additional early peak within the first 10%. Participant 2 recorded a peak at 36%. Participant 3 demonstrated two peaks, one occurring early at 8% and another at 50%. Participant 4 recorded a single peak at 48%.

A summary of peak EMG timings across participants and conditions is presented in Table 7 and illustrated in the bar chart in Figure 29. Note, unavailable data for Participants 3 and 4 during the Fabric EMG sensor trial (trial 2). A calibration instability in the Vicon optical system subsequently produced inconsistent marker tracking, rendering later recordings unusable. As the equipment required manufacturer servicing, the remaining sessions could not be completed within the approved data-collection period. The participants were unable to return for repeat testing due to COVID 19 restrictions.

Participant	Peak EMG recorded (% of gait cycle)		
	Delsys Trigno only	Fabric EMG sensor	Fabric EMG sensor and AFO
1	48.9%	41.6%	45%
2	34% *	40%	36%
3	25%	N/A	8% / 50% *
4	60%*	N/A	48%

*Table 7 - Peak recorded EMG values for each participant with Wireless Delsys Trigno only, fabric EMG sensor and fabric EMG sensor plus AFO. N/A shows trials not successfully recorded due to technical issues in Vicon. * indicates recorded trials showing abnormal*

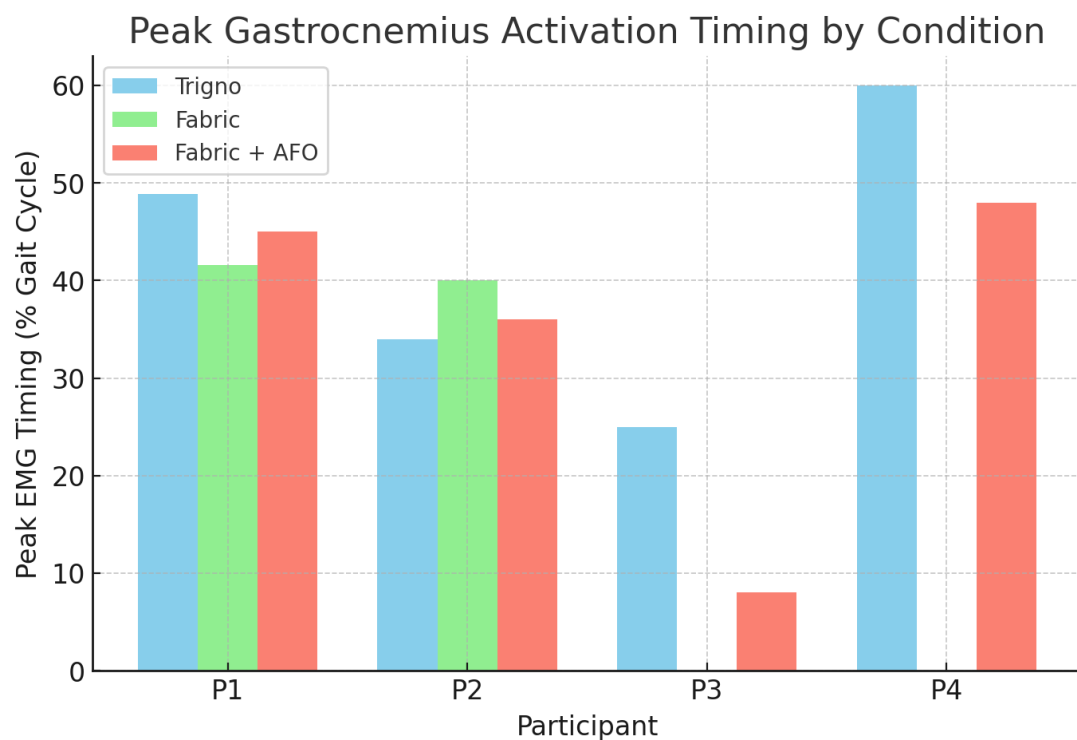


Figure 29 - Bar chart showing peak gastrocnemius activation timing across participants and conditions (Trigno, Fabric, Fabric + AFO).

Descriptive comparison of fabric and Trigno data showed small differences for Participants 1 and 2 (-7.3% and $+6.0\%$, respectively). When the fabric sensor was worn with an AFO, differences relative to Trigno were -3.9% for Participant 1, $+2.0\%$ for Participant 2, -17.0% for Participant 3, and -12.0% for Participant 4. These results indicate that while the fabric sensor broadly captured activation events in line with Trigno, variability increased when used with an AFO.

Figure 30 depicts a bar chart showing the differences in peak EMG timing between the fabric sensors (with and without AFO) and the Trigno system for each participant.

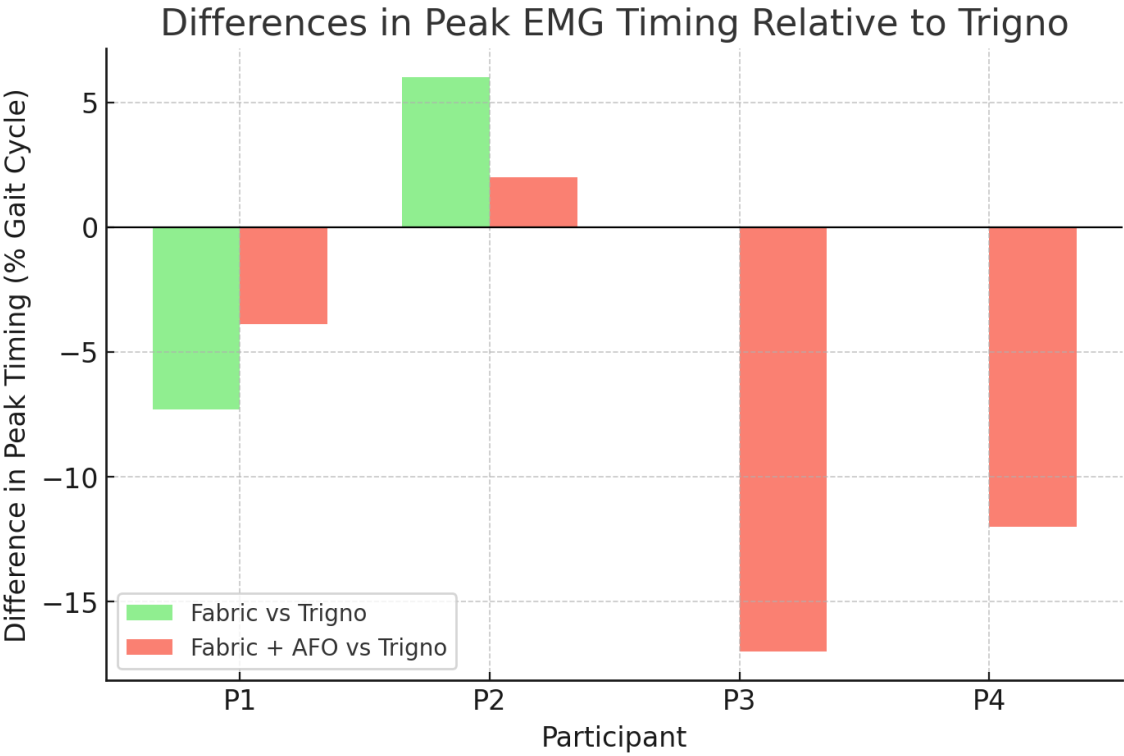


Figure 30 - Bar chart showing the differences in peak EMG timing between the fabric sensors (with and without AFO) and the Trigno system for each participant.

All EMG graphs shown in Figure 31 overpage.

Due to small size, the full graphs can be viewed in Appendix C, section C.4.

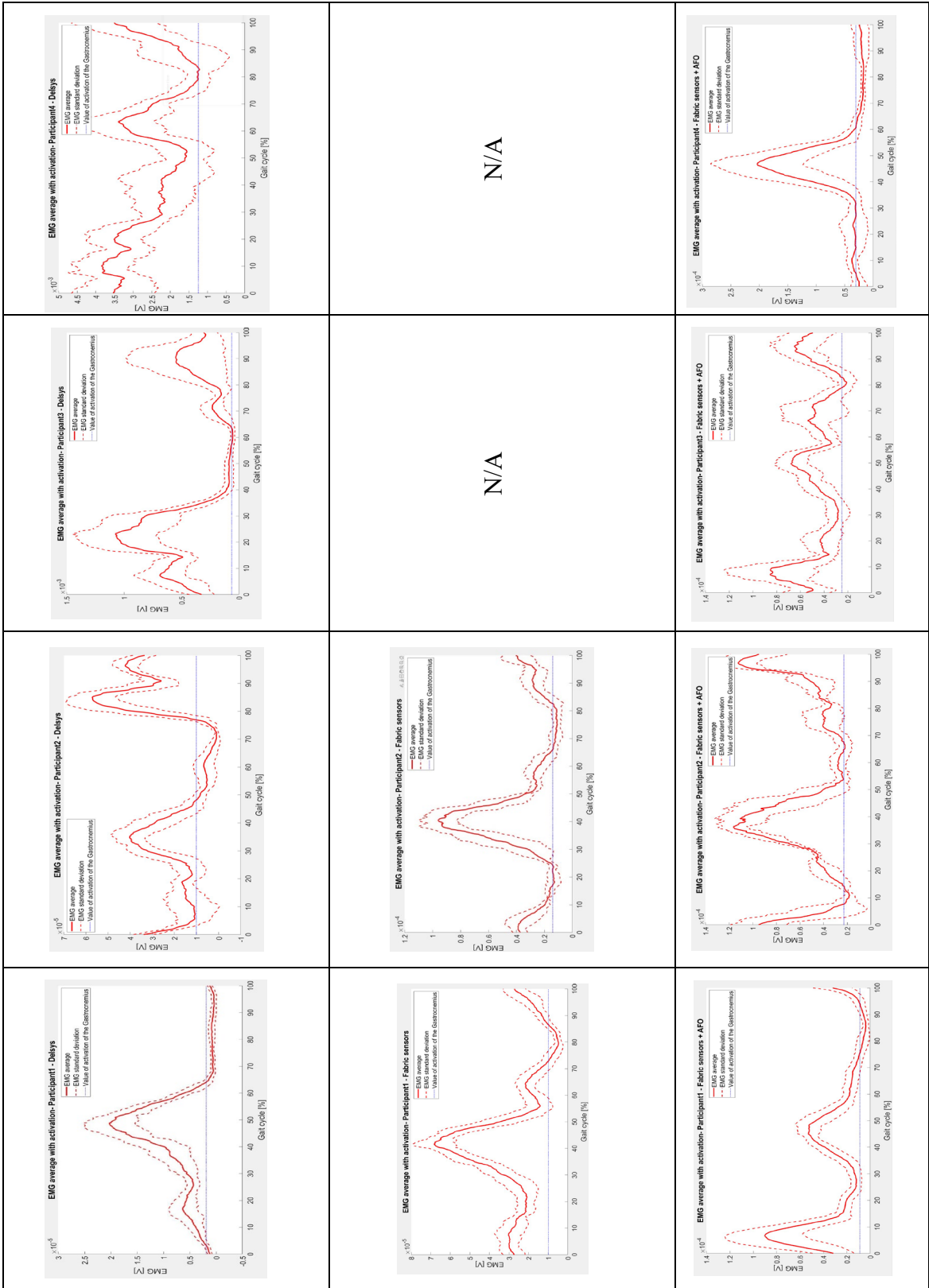


Figure 31 - All available EMG (lateral head Left gastrocnemius muscle) for participants 1 to 4 at each data collection - Delsys Trigno EMG, Fabric EMG and Fabric EMG plus AFO.

7.6 Discussion

7.6.1 activPAL versus Vicon for Peak SVA

The secondary aim of this study was to determine whether the activPAL monitor could provide an acceptable alternative to deriving shank-to-vertical angle (SVA) from Vicon data. While activPAL outputs aligned with Vicon at comfortable and fast speeds, systematic underestimation was evident during slow walking, where errors exceeded the predefined threshold for acceptability.

Agreement analyses added depth to these findings. Bland–Altman plots illustrated the range of variability between systems, with wide limits of agreement at comfortable speed, systematic negative bias at slow speed, and closer alignment at fast speed. Coefficient of variation (CV) analysis confirmed these patterns, with Vicon demonstrating consistently stable measures across speeds ($CV < 0.015$), while activPAL produced higher variability ($CV \approx 0.07–0.09$). These descriptive analyses suggest that activPAL may be most reliable at faster walking speeds, but its accuracy decreases substantially when gait slows.

The unexpected deterioration at slow speed contrasts with expectations that accelerometer-based drift would increase during faster, more dynamic movement (Pedley, 2013). Instead, activPAL's relatively low sampling frequency (20 Hz) appears to be a critical limitation, as comparable low-cost IMUs operating at 100–128 Hz have demonstrated stronger agreement with Vicon (de Jong et al., 2020; Nguyen et al., 2020). Fewer data points per gait cycle likely limit the resolution of subtle kinematic changes, producing systematic underestimation during slower ambulation.

Although activPAL has not been previously validated for SVA estimation, the discrepancies observed here highlight the need for caution in adopting the device for this purpose. Sampling rate appears central to performance, and future studies may need to prioritise higher-frequency IMUs for accurate SVA measurement in clinical gait analysis.

7.6.2 Electromyography (EMG)

The primary aim of this study was to validate the Footfalls fabric EMG sensor against the Delsys Trigno gold standard, including testing under AFO conditions. Results highlighted both areas of agreement and variability across participants.

With the Trigno sensor, peak gastrocnemius activation typically occurred within the expected 40–45% of the gait cycle (Hof et al., 2002; Buurke, Lamothe, van der Woude, & Rob den Otter, 2016). Participant 1 consistently demonstrated peaks within this range across all phases (48.9% Trigno; 41.6% Fabric; 45% Fabric + AFO), aligning most closely with normative values and suggesting stable recording across devices. By contrast, Participants 2–4 displayed less consistent profiles. Participant 2 recorded a primary Trigno peak at 34% with secondary late-cycle peaks at 85% and 95%, while fabric peaks were closer to expected values (40% and 36%). Participant 3 demonstrated early (8–25%) and late-cycle (50–90%) activations across devices, while Participant 4 showed a late Trigno peak at 60% and a Fabric + AFO peak at 48%, alongside unusually high early-cycle activity (4.6×10^{-3} V).

Comparisons of peak timing revealed small differences between Fabric and Trigno in Participants 1 (−7.3%) and 2 (+6.0%), while larger discrepancies emerged when the fabric sensor was worn with an AFO: −3.9% (P1), +2.0% (P2), −17.0% (P3), and −12.0% (P4). These findings were clearly visualised in

bar plots of relative timing differences, where Fabric-only data aligned more closely with Trigno, while AFO use increased variability.

Notably, Participant 4's Trigno data demonstrated abnormally high activity during the first half of the gait cycle, suggesting potential movement artefact or protocol-related error. Previous research has similarly noted that device placement, attachment materials, and orthosis design can influence EMG recordings, particularly when early-cycle peaks coincide with initial contact or loading response (Ohata, Yasui, Tsuboyama, & Ichihashi, 2011).

Ensemble waveform plots further demonstrated that the fabric sensor reproduced the overall shape and timing of gastrocnemius activation but with increased variability in secondary peaks and early-cycle activations. The early peaks observed in Participants 1 and 3 during the first 10% of the gait cycle are consistent with motion artefact from sensor–skin–AFO interactions, as reported in previous orthotic EMG studies (Ohata et al., 2011).

The observed abnormalities in Participants 3 and 4 are more likely linked to trial protocol and user-related variability than to device limitations, given that discrepancies also occurred in the Trigno recordings. This highlights the importance of refining methodological procedures (e.g., electrode placement, securing straps, familiarisation with AFOs) before clinical deployment.

7.6.3 EMG

The primary aim of this study was to validate the Footfalls fabric EMG sensor (with Delsys snap-lead extension) against the Delsys Trigno gold standard. This was undertaken with four healthy participants under three conditions: Trigno only, fabric EMG sensor, and fabric EMG sensor worn with an AFO. Although care was taken to select a posterior leaf spring (PLS) AFO design

representative of standard clinical devices, it should be noted that the participants had no underlying gait impairments and were therefore unfamiliar with orthotic use. This is relevant given previous evidence that AFO stiffness, structural integrity, and trimline design differ substantially between off-the-shelf and custom devices, and these differences can alter gait dynamics (Harper et al, 2014; Totah, Menon, Jones-Hershinow, Barton, & Gates, 2019).

Normative data suggest that peak gastrocnemius activity occurs at ~40–45% of the gait cycle (Hof, Elzinga, Grimmius, & Halbertsma, 2002; Buurke, Lamoth, van der Woude, & Rob den Otter, 2016). Within this cohort, only Participant 1 consistently demonstrated activation patterns within this expected range across all conditions (48.9% with Trigno; 41.6% with fabric EMG; 45% with fabric EMG + AFO). This suggests that, in at least one case, the fabric device was able to replicate established normative values.

In contrast, Participants 2–4 demonstrated variable and less consistent activation profiles. Participant 2 displayed an early Trigno peak at 34% with additional late-cycle activity (85% and 95%), although the fabric recordings were closer to normative values (40% and 36%). Participant 3 recorded multiple peaks, with activity in early stance (8–25%) and again in late stance (50–90%), while Participant 4 demonstrated a late-cycle Trigno peak at 60% and a Fabric + AFO peak at 48%, alongside unusually high activity during early stance (4.6×10^{-3} V).

Additional peaks detected during the first 10% of the gait cycle for Participants 1 and 3 are likely attributable to motion artefact generated by interactions between the skin, the sensor, and the AFO during initial contact and loading response. Comparable findings have been reported by Ohata et al. (2011) in studies of AFO use with integrated sensors, where attachment method and orthosis design significantly influenced EMG signal quality. The abnormalities observed in Participants 3 and 4 are more plausibly explained by protocol- or

user-related variability than by hardware limitations, particularly as irregularities were also present in the Trigno phase of testing.

A further challenge for validation arises from limitations in amplitude data. While Trigno outputs confirmed abnormal high early-cycle activity in some cases, fabric device voltage outputs were not measurable through the Delsys interface, restricting amplitude comparison. Normalisation of Trigno amplitudes to participant maxima permitted limited descriptive assessment but does not allow for robust cross-device comparison. These issues underscore the importance of refining sensor attachment methods, securing mechanisms, and orthotic integration before progressing to larger-scale clinical studies.

From a translational perspective, the feasibility of using fabric EMG sensors in clinical practice is significant. Data collection in rehabilitation settings is often constrained by limited time and resources, leading many clinicians to rely on observational assessments despite known inaccuracies; Hancock et al. (2018), for example, reported up to a 10% minimum significant difference when comparing visual estimates to instrumented goniometry. By contrast, wearable EMG technologies and low-cost digital assessment tools could enable more precise and timely gait analysis, supporting orthotic tuning and rehabilitation planning. As suggested by Nguyen et al. (2020), smartphone-based and wearable approaches offer a practical route to extend gait assessment beyond specialist laboratories, providing clinicians with real-time, context-relevant data.

It must be recognised however, that the cohort sample size in this study cannot contribute significantly towards any confidence. The trend toward functionally useable data with positive comparison against established data collection methods is perhaps the only available conclusion.

The significance of this work lies in demonstrating the feasibility of fabric-based EMG technology to record muscle activity in combination with an orthosis - an application not possible with conventional electrodes due to positioning and electrode size. While the current study identified variability and artefact, the findings indicate a clear pathway for refinement and further validation.

Continued development of this device, coupled with integration into clinical settings, could enhance the precision of gait assessment, reduce reliance on subjective observation, and ultimately support the thesis aim of advancing wearable, orthosis-compatible technologies for stroke rehabilitation.

7.6.4 Critical Reflection and Links to Thesis Aims

This proof-of-concept study addressed two central objectives: the pilot-validation of a novel fabric EMG sensor and evaluation of the activPAL monitor for estimation of shank-to-vertical angle (SVA). Together, these investigations provide important insights into the feasibility of low-cost wearable technologies for gait assessment and their potential integration into stroke rehabilitation.

This study was designed as a proof-of-concept case study, reflecting the influence of external environmental and practical constraints as previously discussed. In contrast, a fully powered validation study would require approximately 46 ambulant stroke participants, with recruitment of around 55 to account for attrition and incomplete datasets.

For the EMG component, the study demonstrated that the Footfalls fabric sensor, with Delsys snap-lead extension, could detect gastrocnemius activation timing in line with normative expectations, including during use with an ankle-foot orthosis (AFO). Although artefacts and variability were evident across participants, the ability to capture meaningful activation profiles through an orthosis represents a significant step forward, since conventional EMG

electrodes cannot provide reliable in-brace data. These findings suggest that further refinement of sensor attachment, orthosis integration, and clinical usability could position fabric EMG technology as a practical solution for in-clinic and real-world monitoring of gait. For the SVA analysis, activPAL outputs aligned with Vicon under comfortable and fast walking conditions but showed systematic underestimation at slow speeds, with mean errors exceeding the acceptable threshold.

Overall, activPAL demonstrated greatest reliability at higher walking speeds but was limited by its low sampling rate of 20 Hz compared to alternative IMUs. These technical constraints highlight that while activPAL may provide value for certain clinical applications, alternative higher-frequency devices are likely required for robust SVA estimation across diverse gait conditions.

When considered together, these findings highlight both the opportunities and challenges of translating wearable sensors into stroke rehabilitation. The EMG study demonstrated feasibility of novel technology for in-brace monitoring, addressing a long-standing limitation in orthosis assessment, while the SVA analysis underscored the importance of careful device selection and validation ensure measurement accuracy.

Both strands reflect the broader thesis trajectory: identifying unmet clinical needs (Chapter 5), developing and prototyping wearable solutions (Chapter 6), and evaluating technical feasibility under controlled conditions (Chapter 7). In doing so, this study reinforces the co-design framework underpinning the thesis. By situating technical development within the realities of clinical practice, including the need for orthosis compatibility, ease of use, and accuracy across conditions, the research ensures that future iterations of these technologies remain aligned with clinical priorities. This alignment is critical not only for successful integration into rehabilitation workflows but also for maximising the potential impact on patient outcomes.

7.7 Summary

This chapter has demonstrated that wearable technologies can provide meaningful insights into gait but also revealed technical and methodological challenges that must be addressed before clinical translation. Validation of the Footfalls fabric EMG sensor confirmed that muscle activation could be recorded both with and without an orthosis, including through in-brace measurement, a capability not possible with current gold-standard systems. Although variability and artefact were observed, these findings highlight the promise of fabric-based EMG as a foundation for orthosis-compatible monitoring tools.

Evaluation of the activPAL wearable monitor showed condition-dependent performance. At comfortable and fast walking speeds activPAL estimates of shank-to-vertical angle (SVA) were broadly acceptable, while systematic underestimation occurred during slow gait. These findings suggest that while activPAL may have value in some contexts, higher-frequency devices are likely required for accurate and consistent SVA measurement.

Together, these results provide proof-of-concept evidence that aligns with the overall thesis aim of advancing accessible gait assessment technologies. Importantly, the chapter identified both the potential of novel wearable systems and the refinements required for successful integration into stroke rehabilitation.

The next chapter builds directly on this work by presenting the final participant study of this body of work, a feasibility study using the third prototype design described in Chapter 6. This progression moves beyond controlled laboratory evaluation to test the practicality and acceptability of the device, providing a critical step toward real-world application.

Chapter 8 – Feasibility study of fabric EMG sensor in novel combination with an Arduino IMU

This chapter reports on the second participant study, which explored the feasibility of a novel wearable system integrating a fabric-based EMG sensor woven into a sock design, combined with an Arduino-driven IMU and a MyoWare EMG module. Building on the earlier proof-of-concept work, the purpose of this study was to assess the practicality and potential of this combined approach for capturing gait-related signals in individuals post-stroke. Consistent with the overarching thesis aims, the study was designed to examine innovative tools that may enhance both the functional and descriptive analysis of gait in general, and stroke rehabilitation in particular.

Acknowledgements – I would like to thank Footfalls and Heartbeats for providing the fabric EMG devices and Becker Orthopaedic for providing the AFOs used during this study.

8.1 Justification of Development Decisions from Chapter 7

8.1.1 Adaptation to full sock design

As outlined in Chapter 6, iterative prototyping led to a shift from discrete fabric EMG electrode patches to a tubular module, and ultimately to a full compression sock format (developed in collaboration with Footfalls & Heartbeats) (see Figure 35). This transition addressed the recurring issues of electrode stability, motion artefact, and wearer comfort identified during laboratory testing. The tubular configuration improved retention compared to patches but revealed limitations during extended wear, including proximal and distal fabric migration, electrode crosstalk, and increased localised pressure. These findings directly informed the decision to adopt a clinically validated Class 1 graduated compression sock as the base garment, which provided stable electrode positioning while ensuring safety for stroke-affected populations with circulatory or sensory comorbidities.

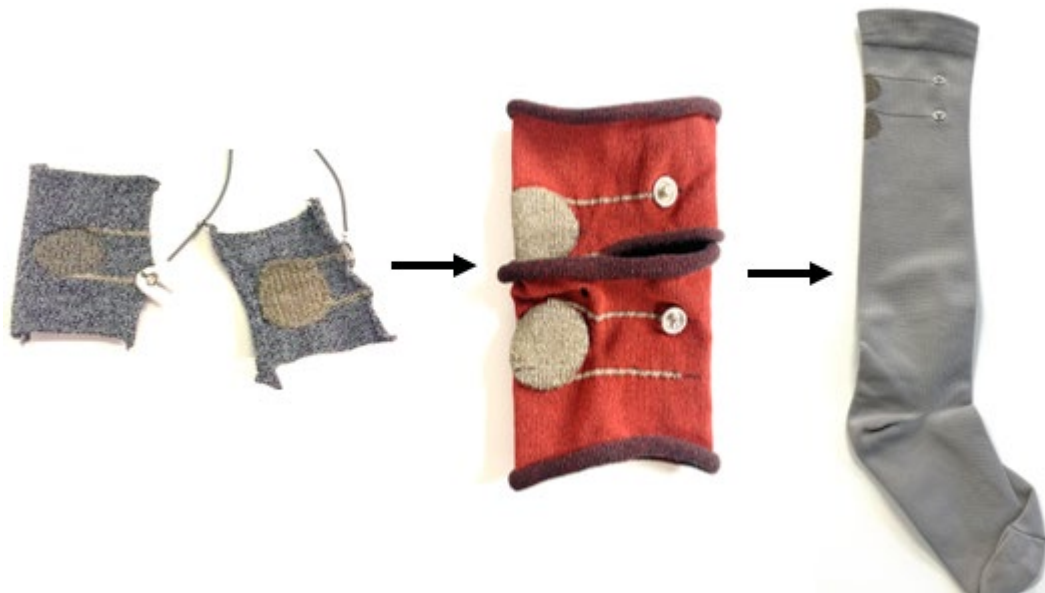


Figure 32 - Illustrating pathway from individual electrode patches to a fully integrated sock design

199

The Arduino Pro Mini platform was selected as the microcontroller due to its low cost, open-access coding environment, and modular compatibility with the MyoWare EMG board and the IMU sensor. Unlike more established wearable IMU systems such as the activPAL4, which had demonstrated validity against Vicon in prior research, the Arduino-based IMU had no existing validation in a clinical gait context. This absence of precedent necessitated preliminary testing to establish reliability and ensure that the simplified, low-cost system could provide biomechanically meaningful outputs.

Accordingly, a pilot validation study was undertaken to compare the Arduino IMU against the Vicon Nexus 3D motion analysis system, considered the gold standard in gait kinematics (Ugbolue et al., 2013; Lee et al., 2023). The Wolfson Gait Laboratory at the University of Strathclyde provided the infrastructure for this comparison, including 12-camera Vicon T-Series cameras operating at 100 Hz.

Retroreflective markers applied using the Plug-in-Gait model enabled sagittal plane shank-to-vertical angle (SVA) calculations, which served as the benchmark against which IMU-derived estimates were evaluated.

This staged validation was critical to the translational pathway of the device: it not only established the construct validity of the Arduino IMU for detecting shank orientation and heel-strike surrogates but also confirmed the feasibility of integrating low-cost, modular hardware into a clinically relevant wearable system.

8.1.2 Vicon comparison

Owing to the absence of prior comparative data evaluating the Arduino Mini IMU against the Vicon motion capture system, an initial pilot validation was

undertaken to establish the preliminary validity of the novel sensor configuration before progressing to data collection with research participants. Conducting this validation phase aligns with good research practice, particularly given the subsequent inclusion of a stroke-affected cohort, where system reliability and data integrity are critical. Two participants were recruited from the University of Strathclyde in accordance with the inclusion and exclusion criteria defined for the main study (see Section 8.4.2).

The objectives of this initial phase were twofold: (1) to evaluate the performance of the bespoke MATLAB processing scripts, developed to filter and transform raw IMU and Vicon outputs into a directly comparable format, and (2) to test and refine the device setup and data acquisition procedures prior to adopting the full study protocol.

8.1.2.1 Measurement systems and expected signal characteristics

The Arduino mini IMU was mounted on the anterior aspect of the shank, with its long axis approximately aligned to the tibial shaft. Raw tri-axial accelerometer and gyroscope signals were zero-lag low-pass filtered at 20 Hz (forward–backward Butterworth). A 20 Hz cut-off is standard for shank kinematics during level walking: it preserves stance-phase content and event dynamics while attenuating soft-tissue artefact and high-frequency noise produced at heel strike.

Biomechanically, the SVA (angle between tibia and vertical in the sagittal plane) rises from loading response through mid-stance as the shank progresses over the foot, reaches a late-stance peak, and then falls entering swing. In both systems we therefore expect a monophasic rise to a single peak in the tens of degrees, with timing in late stance. Small systematic offsets between systems are plausible because coordinate definitions and sensor mounting frames inevitably differ.

8.1.2.2 Methods: acquisition, kinematics, and statistical design

Two healthy adult participants were recruited from the University of Strathclyde community and provided written informed consent prior to testing. Both individuals met the inclusion and exclusion criteria established for the main study (see Section 8.4.2). Testing was conducted in the Wolfson Laboratory under controlled environmental conditions using an overground walking protocol. Each participant completed three 5 m walking trials at a self-selected comfortable speed, allowing replication of the procedures intended for subsequent data collection.

The Arduino Mini IMU was mounted securely on the anterior aspect of the right shank, aligned with the tibial crest and fixed using elasticated medical tape to minimise skin motion artefact. Reflective markers were placed according to the standard lower-limb Plug-in Gait model to enable 3D kinematic tracking using the Vicon Nexus motion capture system (Vicon Motion Systems Ltd., Oxford, UK). Vicon data were sampled at 100 Hz.

After 20 Hz zero-phase low-pass filtering of both accelerometer and gyroscope channels, a brief quiet-standing segment provided an orientation baseline and a simple soft-iron/placement bias mitigation. Heel strike and toe-off surrogates were identified from the sagittal shank angular velocity (prominent zero-crossings with amplitude and refractory-interval criteria). Strides were segmented and time-normalized to 101 samples (0–100 % gait cycle, %GC), ensuring direct point-wise comparison to the Vicon curves.

SVA was derived with a minimal, well-established IMU pipeline:

1. Orientation estimation.

A complementary/Madgwick filter fused low-frequency accelerometer orientation (gravity) and high-frequency gyroscope integration to produce the quaternion $q_s \rightarrow g$ that maps sensor frame to a gravity-aligned global frame (*global + z^ points upward*).

2. Tibial axis in global coordinates.

The IMU axis adhered to the tibial long axis (e.g., sensor y^s) was rotated into global: $\hat{t}_g = q_{s \rightarrow g} [0, \hat{y}_s] q_{s \rightarrow g}^{-1}$.

3. Sagittal projection and angle to vertical.

The tibial axis was projected into the sagittal plane:

(AP-vertical), $\hat{t}_{\text{sag}} = \text{normalize}([t_x, 0, t_z])$.

The SVA was then $\theta = \arctan 2(\hat{t}_{\text{sag}} \cdot \hat{x}_{\text{AP}}, \hat{t}_{\text{sag}} \cdot \hat{z})$ in degrees, adopting a sign convention in which positive angles reflect forward tibial inclination. A short neutral standing pose zeroed the baseline.

This approach requires no multi-sensor fusion across segments: a single sensor suffices because SVA is defined with respect to global vertical, not the thigh or foot.

8.1.2.3 Statistical design and rationale

With the small sample size, the focus was exploratory with a focus on descriptive agreement rather than inferential group testing.

The following were therefore reported:

- Shape similarity by Pearson's r (sensitive to waveform morphology, insensitive to constant offsets).

- Absolute error by RMSE and MAE in degrees, plus normalized RMSE (nRMSE) relative to the Vicon peak-to-peak amplitude to contextualize magnitude.
- Temporal/shape similarity by Dynamic Time Warping (DTW) to reflect residual phase differences.
- Agreement to the normative band as the proportion of the gait cycle within ± 1 SD and ± 2 SD of Vicon.
- Feature comparators: the difference in peak SVA (deg) and difference in peak timing (points of %GC).
- Bland–Altman plots of device–Vicon differences versus average amplitude to visualize bias and dispersion across the signal range.

These metrics are standard, together giving a rounded view of (i) whether the wearable reproduces the shape of the canonical SVA curve, (ii) how large its errors are in clinical units (degrees), and (iii) where any residual deviation concentrates in the cycle. Because time-normalized samples are highly autocorrelated, Bland–Altman is interpreted descriptively.

8.1.2.4 Results

Participant A

The Arduino IMU reproduced the canonical SVA morphology convincingly: a clean rise through stance, a single late-stance peak, and a decline into swing. Quantitatively, shape similarity was strong ($r = 0.74$). Absolute errors were moderate (RMSE = 9.60° , MAE = 7.56°), which corresponds to nRMSE =

21.30 % of the Vicon dynamic range—consistent with a wearable capturing the bulk of the kinematic pattern while diverging in specific regions.

Agreement to the population band was high: 77.2 % of the cycle lay within ± 1 SD, and 98.0 % within ± 2 SD of Vicon—indicating that the device waveform generally sits inside normal variability, with only focused departures. The mean bias was small and slightly negative (-2.22°), i.e., the IMU reported on average a marginally lower SVA than Vicon. The DTW distance (334.3) confirmed a modest phase mismatch: the principal peak in Participant A occurred earlier than Vicon's (-22 points of %GC) and was smaller by $\sim 5^\circ$ (peak difference = -5.25°).

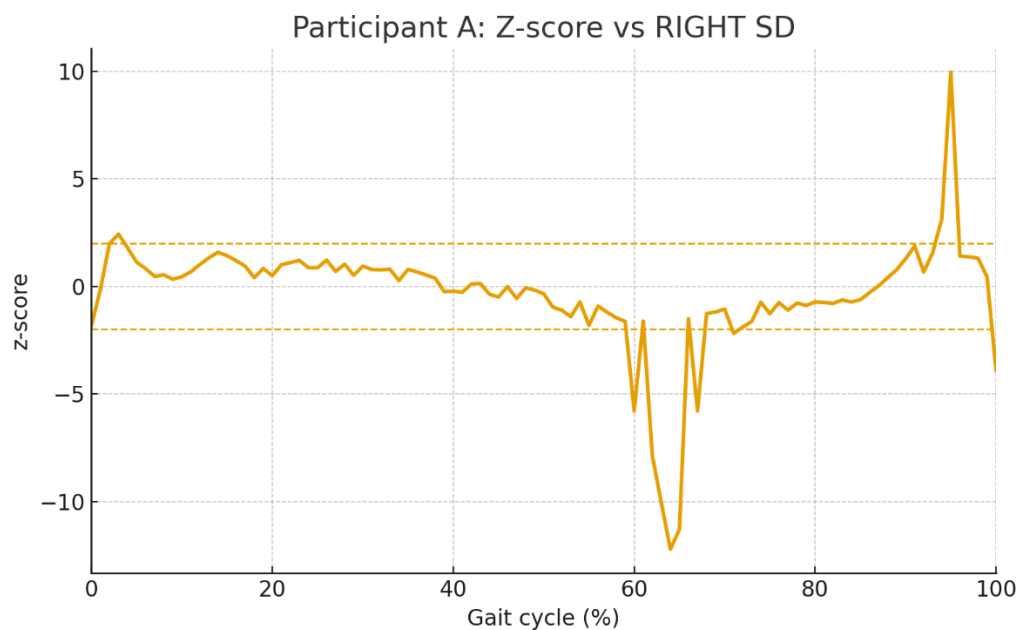


Figure 34 - Participant A: Z-score vs RIGHT SD

The difference and z-score traces (see Figure 34) localised these deviations: during early to mid-stance, the device modestly overshot the mean waveform band, and around late stance it ran slightly below, with transient $|z|$ values approaching but rarely exceeding 2.

A more pronounced deviation was observed around 65 % of the gait cycle, corresponding to the transition from terminal stance into pre-swing. This feature likely reflects a brief period of increased segmental acceleration, during which inertial and gravitational components become more difficult to separate, leading to short-duration orientation error. Bland–Altman analysis (Figure 35) showed differences scattered narrowly about a near-zero bias across the amplitude range, consistent with the small mean bias reported above.

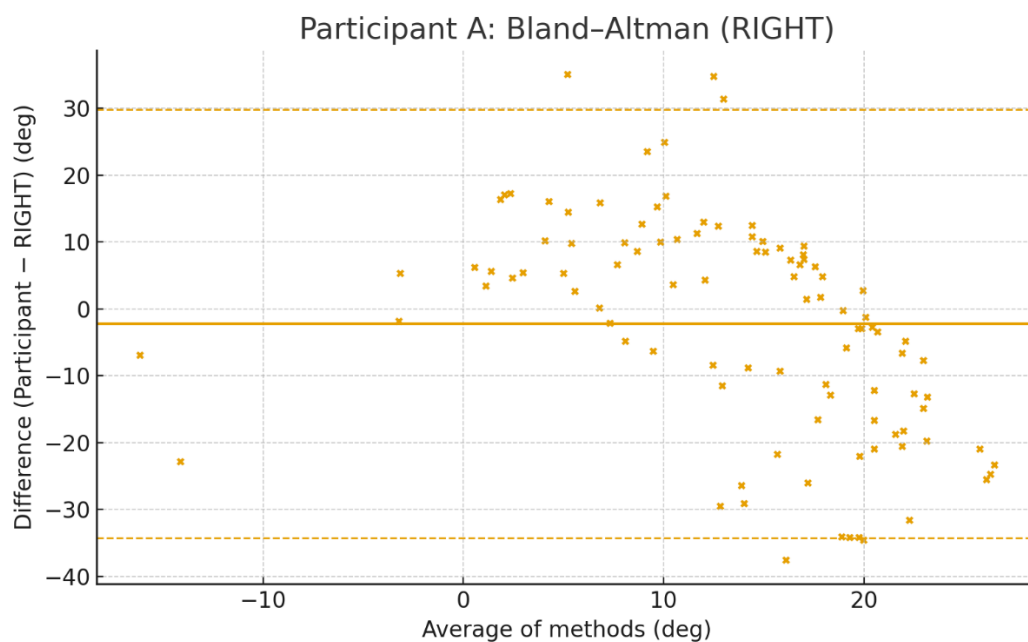


Figure 35 - Participant A: Bland–Altman plot for Right side

Interpretation

For a single-sensor, single-segment estimate, Participant A's right-leg SVA aligns closely with Vicon both in form and magnitude. Residuals concentrate at the peak, earlier timing and slightly reduced amplitude, consistent with small differences in event detection and coordinate realization. For research

purposes the waveform is fit-for-purpose; for strict interchangeability at the feature level, a light calibration targeting the peak region would likely eliminate most of the remaining error.

Participant B

Participant B exhibited even stronger shape agreement ($r = 0.812$). Absolute errors were slightly lower than Participant 61 ($\text{RMSE} = 8.97^\circ$, $\text{MAE} = 6.87^\circ$) with $\text{nRMSE} = 20.2\%$. A larger proportion of the cycle lay inside the normative variability: 86.1% within ± 1 SD and 99.0% within ± 2 SD. The device showed a small, consistent negative bias (-5.00°), i.e., a tendency to read below the Vicor RIGHT mean. The DTW distance (282.7) and features again indicated a phase lead and smaller peak: peak difference -10.36° , peak timing -13 points of %GC, concentrated around late stance.

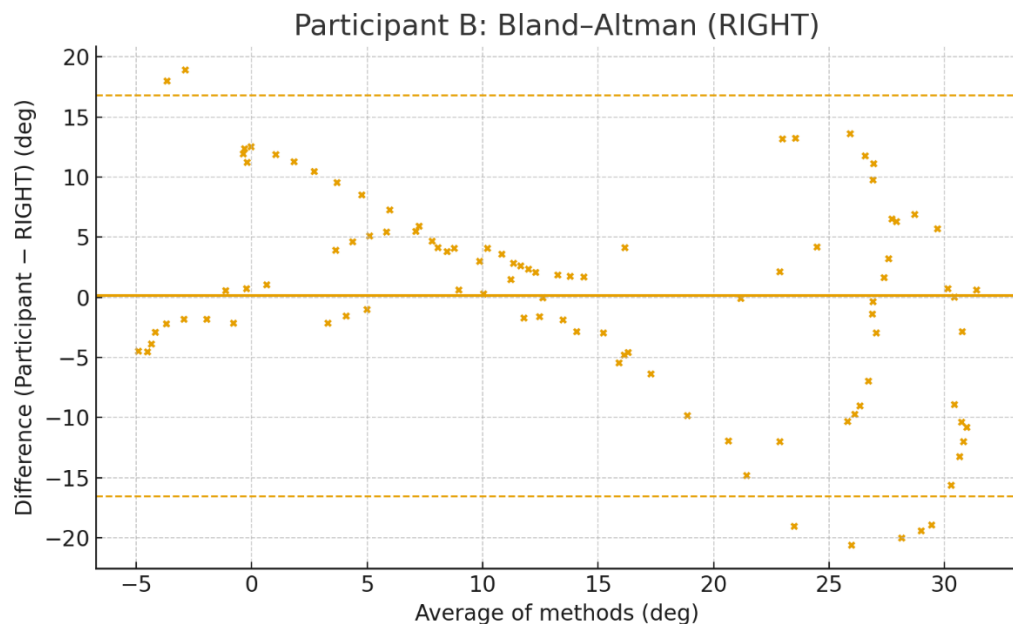


Figure 36 - Participant B: Bland-Altman plot

Interpretation

Participant B demonstrates robust convergence to Vicon across the entire gait cycle, with very high in-band agreement and only limited, interpretable deviations at the late-stance peak. For this participant, a simple constant offset would neutralise most of the small bias; the chief residual would be peak timing, which is an expected consequence of using a single-sensor tibial axis and event surrogates.

Bland–Altman analysis for Participant B (vs Vicon) (see Figure 36) showed negligible mean bias ($+0.16^\circ$) with 95 % limits of agreement from -16.5° to $+16.8^\circ$. However, the difference clearly increased as the average SVA angle increased, evidenced by a significant negative association with the mean (slope ≈ -0.30 deg/deg). This indicates a proportional bias, whereby larger SVA values during late stance were slightly underestimated relative to Vicon.

Such proportional bias likely reflects the increasing influence of segmental angular acceleration and alignment drift at higher angles, when the shank approaches its maximal inclination and inertial components dominate the accelerometer signal. For the current system, this pattern suggests that while low- to mid-range angles are captured with high fidelity, further refinement of sensor calibration and fusion weighting may be necessary to maintain accuracy in the terminal stance phase, where dynamic acceleration is greatest.

Despite these localised deviations, the overall correspondence between the Arduino-based IMU and Vicon is sufficiently strong to justify progression to the next testing phase. Correlation coefficients above 0.7, RMSE values near 9° , and near-zero mean bias indicate that the system achieves acceptable agreement for gait-related applications. Most importantly, waveform shape and timing are preserved across the cycle, supporting its suitability for detecting relative condition-dependent differences, for example, between AFO-restricted

and unrestrained walking, rather than for absolute angular quantification. With the application of a simple constant or proportional correction, and continued attention to sensor alignment and synchronisation, the system demonstrates adequate validity for use in the forthcoming stroke-affected cohort testing.

Comparison vs Vicon	r	RMSE (°)	nRMSE (% range)	MAE (°)	Mean bias (°)	DTW	% within ±1 SD	% within ±2 SD	Peak diff (°)	Peak timing diff (pts)
Participant A	0.735	9.60	21.3	7.56	-2.22	334.3	77.2	98.0	-5.25	-22
Participant B	0.812	8.97	20.2	6.87	-5.00	282.7	86.1	99.0	-10.36	-13

Figure 37 - Agreement for right-leg SVA (0–100% GC): Arduino anterior-shank IMU vs Vicon Nexus—r, RMSE/MAE, nRMSE, bias, DTW, %±1/±2 SD, peak diff & timing.

Comparison

Across both participants, three patterns were consistent. First, waveform shape was captured well ($r \approx 0.74\text{--}0.81$), reinforcing that a single anterior-shin IMU can provide a valid qualitative and semi-quantitative representation of SVA.

Second, absolute errors were modest ($\text{RMSE} \approx 9^\circ$), indicating that IMU-derived angles closely tracked the Vicon reference across the gait cycle. Most of each cycle fell within ± 1 SD of Vicon's mean waveform, and nearly all data points lay within ± 2 SD, suggesting good temporal and magnitude correspondence.

Third, residual discrepancies occurred primarily at specific gait-cycle features: IMU-derived peaks were typically earlier and smaller, reflecting expected differences arising from single-segment orientation estimation and stride-normalisation procedures rather than true measurement bias.

An RMSE of approximately 9° falls within the range typically reported for single-shank IMU validation studies, which commonly demonstrate mean absolute errors between 5° and 10° when compared with optical motion capture (Takeda et al., 2016; Robert-Lachaine et al., 2017; Zhang et al., 2019).

Given that SVA itself exhibits a natural inter-trial variability of roughly $6\text{--}8^\circ$ in healthy gait (Ounpuu et al., 1991), this level of agreement indicates that the IMU provides clinically acceptable accuracy for assessing shank alignment in functional contexts.

8.1.2.5 Discussion

The results show that a minimal hardware configuration, a single Arduino-class IMU filtered at 20 Hz, can reproduce the canonical SVA morphology with small mean bias and moderate absolute error. The high proportion of samples within the Vicon ± 1 SD band indicates that the device's output is not merely qualitatively similar but quantitatively aligned with the population norm over most of the gait cycle. This is precisely the performance envelope typically sought for longitudinal monitoring and comparative profiling in clinics and field studies, where portability and repeatability often outweigh the marginal precision gains of optical systems.

Peak SVA is a localised feature shaped by three influences: (i) phase detection (how %GC is defined stride-to-stride), (ii) coordinate realization (IMU tibial axis vs Vicon segment definition), and (iii) soft-tissue artefact near late stance when acceleration and angular velocity are largest. A single shin sensor will inevitably register a slightly different phase landmark than multi-segment optical models.

That the observed differences are limited to the peak zone—with the rest of the cycle closely tracking the Vicon mean—strongly suggests construct validity: the device sees the same underlying biomechanics and deviates only where measurement definitions diverge most sharply.

These findings support the Arduino IMU as a justifiable comparison to Vicon for research-grade assessment of SVA, appropriate for trend analysis, between-condition comparisons, and within-subject monitoring. Where precise numerical interchangeability is required, the remaining discrepancies are small, predictable, and localizable (peak magnitude and timing), and can be reduced further by minimal calibration without compromising the methodological simplicity that makes the wearable attractive in the first place.

8.1.3 Justification of Final Device Design

In conclusion, adopting a full graduated-compression sock as the EMG carrier was decisive for data quality and clinical usability. The sock eliminated the electrode migration, crosstalk, and comfort issues observed with patches and tubular sleeves, while offering a safe, repeatable interface for stroke populations. Because EMG alone did not provide sufficiently reliable gait-event timing, embedding a thin heel contact sensor and a single anterior-shank IMU gave the EMG signals the temporal and kinematic context required for valid interpretation, without compromising wearability.

The Arduino-class IMU, filtered at 20 Hz, proved technically adequate for shank-to-vertical angle (SVA) estimation and compared favourably with Vicon. Across both preliminary participants, residual differences were localized to the peak region (earlier, slightly smaller peaks), consistent with expected event-timing and coordinate-definition effects and amenable to minimal calibration if strict numerical interchangeability is required.

Taken together, these findings justify the final design: the full-sock EMG delivers mechanically stable, clinician-acceptable sensing, and the Arduino-based IMU provides reliable, research-grade SVA that closely tracks Vicon across the gait cycle. The system therefore meets the practical and scientific requirements for longitudinal monitoring, condition comparisons, and clinical translation, while retaining the affordability and openness needed for broad deployment.

8.2 Introduction

The development and validation of the fabric-integrated EMG sensor combined with an Arduino-based inertial measurement unit (IMU) has, to this point, been conducted in controlled environments and with non-affected participants (see Chapter 7). This staged approach was deliberately adopted to ensure technical feasibility, reliability of signal acquisition, and the identification of design flaws prior to deployment in the stroke population. The present study therefore represents a critical step in establishing whether this novel system can be applied in real-world contexts with stroke survivors, for whom variability in gait patterns, muscle activation, and tolerance of wearable devices pose substantial methodological and clinical challenges.

Existing gold-standard methods, such as three-dimensional gait analysis (3DGA) coupled with laboratory-grade EMG, provide invaluable mechanistic insights but remain largely inaccessible to many stroke services due to cost, expertise requirements, and ecological limitations. Conversely, widely used functional outcome measures (e.g., 10-Metre Walk Test, Timed Up and Go) are pragmatic but cannot distinguish between true neuromuscular recovery and compensatory strategies. By combining EMG with segmental kinematics derived from IMUs, the current device aims to bridge this gap, offering data of mechanistic relevance within a format that is potentially scalable for clinical and community use.

Critically, feasibility in stroke populations cannot be assumed from healthy participant validation. Stroke-affected gait is characterised by asymmetry, abnormal co-contraction, reduced propulsion, and fatigue (see Chapter 2, Sections 2.3–2.4). These factors not only increase the complexity of data interpretation but also raise practical considerations, such as skin–electrode contact over spastic or atrophic muscle, tolerance of the compression-garment

form factor, and the potential interaction with ankle–foot orthoses (AFOs). A rigorous feasibility study is therefore required to examine not only the technical performance of the device in this population but also its usability, safety, and acceptability.

The design of this study follows logically from the iterative action-research framework underpinning the thesis: first identifying clinical need (Chapter 5), then developing and validating the prototype (Chapters 6 and 7), and finally applying it in the intended clinical population. This approach ensures methodological coherence and aligns with translational research priorities, where early demonstration of feasibility and patient acceptability are essential precursors to larger-scale validation or clinical effectiveness trials.

8.2.1 Ethics

All procedures were reviewed and approved by the University of Strathclyde - UEC22/76 Kerr/Campbell. The study adhered to the Declaration of Helsinki (2013 revision) and to institutional policies regarding the recruitment and testing of vulnerable populations. Particular attention was paid to safeguarding stroke survivors, who may present with cognitive or communicative impairments in addition to physical disability. Consent procedures were therefore adapted to ensure comprehension, with information sheets written in accessible language and opportunities for participants to involve carers or family members in the consent process.

Given the experimental nature of the device, a comprehensive risk assessment was conducted. The primary risks identified were potential skin irritation from textile electrodes, fatigue from repeated walking trials, and interaction with prescribed orthotic devices. Mitigation strategies included the use of medical-grade conductive yarn, monitoring of skin integrity pre- and post-testing, scheduled rest periods, and close liaison with participants'

treating clinicians to ensure orthotic compatibility. Data were anonymised and stored in accordance with UK GDPR and University of Strathclyde research governance requirements.

8.3 Aim

The primary aim of this study was to evaluate the feasibility of deploying the novel fabric-integrated EMG sensor and Arduino-based IMU system in individuals post-stroke. Specifically, the study sought to determine whether reliable neuromuscular and kinematic data could be collected in a clinical population where gait is characterised by asymmetry, abnormal co-contraction, and frequent orthotic use.

Feasibility was considered across three domains:

Technical feasibility – the ability of the system to acquire readable EMG and IMU signals in stroke survivors, accounting for spasticity, reduced motor unit recruitment, and potential interference from orthoses.

Practical feasibility – participant tolerance of the wearable device, including donning/doffing, comfort during walking trials, and compatibility with rehabilitation practices.

Analytical feasibility – whether the collected data could meaningfully characterise gait events (e.g., foot contact, swing clearance) and muscle activation timing in line with known post-stroke gait deviations, thereby supporting future mechanistic and clinical interpretation.

A secondary aim was to explore whether the device offered insights beyond those available through functional outcome measures or observational gait

analysis. In doing so, this feasibility study intended to establish the foundation for larger-scale validation and clinical effectiveness trials, and to clarify the role of wearable EMG-IMU systems as a bridge between gold-standard laboratory measures and routine stroke rehabilitation practice.

8.4 Method

8.4.1 Study Design

This feasibility study employed a within-subject observational design, structured to test the performance of the novel fabric EMG–IMU system in both non-affected and stroke-affected populations. The inclusion of a healthy cohort prior to stroke survivor testing was intentional, enabling direct comparison of data quality and signal interpretation across groups while also safeguarding participant safety.

The study was conducted in a controlled gait laboratory setting, ensuring that the novel system could be benchmarked against established instrumentation while minimising environmental confounders. Walking trials were undertaken overground, reflecting ecological validity more closely than treadmill-based paradigms and aligning with the device's intended use in real-world rehabilitation.

The design emphasised feasibility rather than hypothesis testing. Accordingly, sample size was determined pragmatically, reflecting both the exploratory nature of the study and recruitment limitations within the stroke rehabilitation population. Data were analysed descriptively, with emphasis on signal clarity, detection of gait events, and participant tolerance, rather than inferential statistics. This methodological stance is consistent with early-phase feasibility

work in rehabilitation technology development, where the primary objective is to determine whether subsequent hypothesis-driven trials are warranted.

8.4.2 Participants

8.4.2.1 Healthy cohort

A preliminary cohort of non-affected (“healthy”) adults was recruited to establish baseline feasibility and to optimise the signal acquisition protocol prior to testing with stroke survivors. Recruitment was conducted via the University of Strathclyde community, using poster and email advertisements.

Inclusion criteria required participants to be aged ≥ 18 years, with no self-reported neurological, musculoskeletal, or cardiovascular conditions that could influence gait. Furthermore, participants must have no comorbidities affecting their ability to wear a single compression sock device and be able to walk 10 to 15m without rest.

This group was intended to provide reference data for assessing device performance under conditions of unimpaired gait, thereby distinguishing technical artefacts from pathological signal features.

8.4.2.2 Stroke cohort

Stroke survivors were recruited through their involvement with the Sir Jules Thorn Centre for Co-creation of Rehabilitation Technology, where they were already participating in preliminary gamification-based rehabilitation trials. This recruitment pathway provided practical advantages: participants were familiar with research staff, accustomed to testing environments, and motivated to

engage in rehabilitation-focused innovation. These factors were expected to support adherence and tolerance, which are critical feasibility outcomes.

However, this recruitment strategy also introduces potential selection bias. Participants already engaged in co-creation and gamification trials may represent a subgroup of stroke survivors who are more technologically inclined, physically able, and motivated to participate in research compared to the broader clinical population. Such individuals may tolerate wearable devices more readily and demonstrate higher engagement with novel protocols, potentially overestimating acceptability and feasibility. Conversely, their experience of repeated testing may also introduce performance bias, as they may alter their gait or behaviour in response to perceived study expectations.

To mitigate these concerns, eligibility criteria were applied consistently. Participants were required to:

- Have a confirmed clinical diagnosis of stroke with residual gait impairment.
- Be aged ≥ 18 years at time of testing.
- Be able to walk at least 10 metres independently (with or without an ankle–foot orthosis).
- Have sufficient cognitive and communicative ability to provide informed consent.

Exclusion criteria included:

- Comorbid neurological diagnoses (e.g., Parkinson’s disease, multiple sclerosis).
- Current musculoskeletal disorder or injury which could be negatively affected though inclusion in the trial.

- Dermatological conditions precluding safe electrode contact.
- Severe cardiorespiratory impairment preventing safe participation in walking trials.

While these measures reduced heterogeneity, it is acknowledged that the sample may not fully represent the wider stroke population. The implications of this selection bias are addressed further in the discussion (Section 8.6), where feasibility findings are interpreted considering both the opportunities and limitations of recruiting through an engaged, co-creation-oriented subgroup.

Each participant was pre-screened by a registered orthotist to ensure safe AFO fit, assess spasticity and alignment, and confirm musculoskeletal safety. Participants also underwent a structured musculoskeletal assessment of the lower limb, including evaluation of foot and ankle range of motion to identify any restrictions that may influence gait pattern or tolerance to orthotic intervention. The Silfverskiöld test was applied to differentiate between gastrocnemius and soleus tightness, with participants required to achieve neutral ankle position (90°) with the knee extended; those unable to do so were excluded to avoid inclusion of fixed equinus or significant gastrocnemius contracture. In addition, participants were screened for pathological lower limb alignment, including excessive knee hyperextension (*genu recurvatum*), varus/valgus deformity, and foot posture abnormalities, which could alter loading patterns during gait or be exacerbated by participation in the study.

These assessments ensured that individuals with musculoskeletal conditions that could be negatively impacted by walking trials or device use were identified and excluded where appropriate, thereby maintaining participant safety and ensuring that observed gait characteristics were representative of neurological impairment rather than confounding orthopaedic factors.

Longer preparation and recovery times were scheduled, and continuous clinical monitoring was maintained throughout each session. These measures were consistent with recommendations for safe rehabilitation practice in stroke populations (Winstein et al., 2016).

While these criteria do not eliminate selection bias inherent to the recruitment pathway, their consistent application reduces variability in key clinical and functional characteristics rather than bias itself, thereby supporting more comparable interpretation of feasibility outcomes within the defined cohort.

8.4.3 Protocol

Participants attended a single gait laboratory session. The novel fabric-based EMG–IMU sock was fitted over the tested limb. Textile electrodes were integrated over the lateral head of the gastrocnemius in line with SENIAM placement guidelines. The IMU was mounted anteriorly at the shank, chosen for its alignment with sagittal-plane motion and sensitivity to shank-to-vertical angle (SVA), a parameter central to gait and orthotic tuning.

To ensure tolerance, participants were allowed to acclimatise to the device for up to five minutes of walking at self-selected pace, following initial donning. Comfort was evaluated using a visual analogue scale (VAS), recorded at three timepoints: (i) immediately after donning, (ii) after the acclimatisation period when the participant first stood and walked, and (iii) following completion of the testing protocol.

This iterative assessment provided both subjective and temporal perspectives on device acceptability. Participants were asked to verbally confirm readiness before data collection commenced.

8.4.3.1 Walking trials

Each participant completed three walking trials along a five-metre walkway, marked with tape at the start and end points. At the beginning of each trial, a five-second stationary phase was observed to facilitate subsequent data processing and signal segmentation.

During each trial, the researcher used a stopwatch to record the time taken to traverse the walkway, starting when the participant's lead foot crossed the initial line and stopping at the final line. These timings were collected alongside the IMU data to enable calculation of relative gait speed and to provide a pragmatic measure of functional mobility.

To mitigate fatigue and prioritise participant safety, chairs were placed at either end of the walkway, allowing rest as needed between trials. Stroke survivors were encouraged to take breaks to avoid gait deterioration associated with exertion, which could confound signal interpretation.

8.4.3.2 AFO Condition for Stroke-Affected Participants

Where stroke-affected participants habitually used an ankle foot orthosis (AFO), trials were conducted using a GEO off-the-shelf (OTS) AFO rather than their usual custom polypropylene device. This design choice ensured standardisation across participants by controlling for shank inclination and material behaviour - unlike polypropylene, which is prone to material creep and highly variable due to bespoke prescription, the GEO AFO provided a consistent ankle alignment across the cohort.

Additionally, the GEO AFO was considered suitable for use with the stroke-affected cohort because it provided a controllable, repeatable orthotic condition while retaining features representative of clinically used anterior-shell

carbon-composite AFOs. It is acknowledged that the GEO is not a routine first-line prescription in many services, partly because the hinge system increases cost compared with simpler off-the-shelf devices. However, this design was deliberately selected because the hinge mechanism allowed the option of adjustable fixation during protocol development, enabling ankle motion to be constrained or modified without changing the entire orthosis. This was important methodologically, as future testing could explore different ankle configurations while altering the fewest possible variables and preserving comparability between conditions.

Although the GEO itself is less commonly prescribed, its anterior-shell configuration reflects the mechanical principles of commercially available off-the-shelf AFOs such as the Allard ToeOFF family, which are designed to assist foot clearance and provide dynamic stance-phase support through an anterior tibial shell and carbon-composite structure (Allard UK, n.d.). Carbon-composite AFOs have been used in adult stroke populations, with evidence suggesting potential improvements in walking speed and energy cost compared with walking without an AFO (Danielsson & Sunnerhagen, 2004). The anterior-shell design is most appropriate for users without significant uncontrolled knee extension pathology, as its three-point force system primarily supports tibial progression and knee stability during mid- to late stance rather than correcting severe genu recurvatum or complex multi-planar instability. For this reason, suitability for GEO AFO use was screened individually by a registered orthotist prior to testing

While bespoke AFO fabrication and iterative tuning would likely have optimised gait for each participant by aligning device characteristics to individual biomechanics, this approach was not feasible within the study design. Such a process would have required individual casting, manufacture, and repeated review appointments to achieve appropriate alignment and functional outcomes, extending participation over several weeks. This would have

substantially increased participant burden, particularly in a stroke-affected cohort, and raised the likelihood of attrition within an already limited sample. In addition, the financial and logistical demands associated with bespoke orthotic provision were disproportionate to the exploratory, device-focused aims of the study.

As such, a standardised off-the-shelf solution was considered an appropriate compromise, enabling controlled comparison across participants while maintaining feasibility within the constraints of the research.

Suitability for fitting with the OTS AFO was assessed individually by a registered orthotist, ensuring no negative impact of device substitution on participant safety or gait performance. The OTS AFO was set at 90° with a rigid ankle, reflecting a common clinical prescription for stability post-stroke. This configuration was selected for two reasons: (i) to test the sensitivity of the EMG–IMU system under a fixed-ankle condition typical of post-stroke AFO use, and (ii) to evaluate usability when paired with an anterior-shell design, representing a lightweight AFO style increasingly employed in community rehabilitation.

8.4.3.3 Instrumentation

The fabric EMG sock (Footfalls & Heartbeats Ltd., Nottingham, UK) was constructed from a compression textile with embedded conductive yarn stitched into electrode zones. Electrodes were aligned with SENIAM guidelines (Hermens et al., 2000) over the tibialis anterior and gastrocnemius muscles. Signal transfer was achieved via snap connectors integrated into the cuff. The electrodes were linked to an Arduino Mini–based MyoWare circuit (SparkFun SEN-13723, Niwot, CO, USA), which amplified, rectified, and integrated the EMG signal in real time.

A Phidgets force-sensitive resistor (FSR; Phidgets Inc., Calgary, Canada) was embedded in the heel section of the sock to identify stance and swing transitions (Pickford et al., 2020). The wearable system also incorporated an Arduino-driven inertial measurement unit (IMU) comprising a triaxial accelerometer and gyroscope, enabling measurement of angular velocity, shank-to-vertical angle (SVA), and cadence.

All signals (EMG, IMU, FSR) were synchronised and exported as .CSV files for subsequent analysis in MATLAB (R2020a; MathWorks, 2021).

8.4.3.4 Data Collection

Surface EMG signals were recorded at 20MHz, with preprocessing (band-pass filtering, rectification, linear envelope) conducted offline. IMU data were synchronised and processed to extract temporal gait events and SVA profiles.

Stopwatch timings were compared against IMU-derived stride segmentation to evaluate internal consistency. Signal clarity, repeatability, and correspondence with expected gait features were prioritised as key feasibility outcomes.

Finally, participant-reported comfort scores were gathered as detailed in section 8.4.3.1 using a visual analogue scale accessible to a wide population of adults and familiar to the participants due to similarity to pain testing scales regularly used as outcome measures in rehabilitation settings.

8.4.4 Data Analysis

8.4.4.1 EMG processing

Surface EMG signals were initially acquired via the MyoWare board, which incorporates an onboard active first-order high-pass filter at approximately 20.8 Hz and a first-order low-pass filter at approximately 498 Hz (Advancer Technologies, 2025). To align with standard EMG processing for lower-limb muscle activity, the signals were further processed in MATLAB using a sixth-order Butterworth band-pass filter (20–400 Hz) to retain the physiological frequency content while excluding movement artefact and high-frequency noise. A 50 Hz notch filter was then applied to remove mains electrical interference, a common contaminant in indoor laboratory recordings. The filtered signals were full-wave rectified and smoothed using a 200-frame moving-average window (≈ 200 samples), thereby generating the linear envelope required for temporal analysis of muscle activation patterns (De Luca, 1997).

The MATLAB script executed these steps in sequence for each channel: (1) apply the Butterworth band-pass filter (zero-phase via `filtfilt`), (2) apply the 50 Hz notch filter (design via `iirnotch`), (3) rectify the signal (absolute value), and (4) compute the moving-average envelope over 200 samples (using the `movmean` function). The envelope output was then synchronised with gait-epoch markers for subsequent temporal analyses.

Surface EMG waveforms were time-normalised to 0–100% gait cycle (101 points) and amplitude-normalised to a common reference to retain between-group magnitude differences: $\%RefMVC = (value / reference) \times 100$, where the reference was the non-stroke group's mean peak amplitude (see §8.5.4). Participants were grouped as Non-stroke, Stroke, and Stroke + AFO. Feature windows were pre-specified: stance (0–60% GC), push-off (40–60% GC), whole cycle (0–100% GC), and t_{peak} (peak timing, %GC). Group summaries report means and SDs for %RefMVC features.

For inference, non-parametric tests were used throughout given small n and non-Gaussian distributions. A Kruskal–Wallis test assessed differences across the three groups. Mann–Whitney U tests were used for pairwise contrasts with Holm correction for multiple comparisons. Cliff's δ quantified effect sizes for amplitude and timing features, with standard magnitude labels. To explore time-localised differences, point-wise non-parametric tests were run at each %GC sample with Benjamini–Hochberg FDR control ($q \leq 0.05$). All tests were two-sided.

8.4.4.2 IMU processing

Raw accelerometer and gyroscope outputs from the anteriorly mounted IMU were filtered to reduce high-frequency noise before derivation of angular velocity and orientation profiles. SVA waveforms were segmented and time-normalised (101 points). Agreement to reference and between-group differences were evaluated using Pearson's r , RMSE/MAE (and nRMSE), DTW distance, $\pm 1/\pm 2$ SD in-band proportions, and feature differences (peak magnitude and timing).

Agreement/bias was visualised with Bland–Altman plots (interpreted descriptively due to autocorrelation). For group contrasts, four features (peak SVA, t_{peak} , mid-stance 40–60% GC, AUC 0–60% GC) were compared using Mann–Whitney U with bootstrap 95% CIs for Hedges' g , and a 1-D cluster-based permutation test (Welch t across %GC; 5,000 permutations; two-sided $\alpha = 0.05$) controlled family-wise error for time-series inference.

8.4.4.3 Gait event detection and synchronisation

A pressure-sensitive footswitch (FSR) placed under the heel provided an independent reference for gait cycle segmentation. Heel-strike and toe-off

timings derived from the FSR were used to align IMU signals and to synchronise EMG envelopes to stance and swing phases. Bespoke MATLAB code was written to facilitate this function – see Appendix D.3.

This step was critical to ensure that observed activation patterns could be reliably mapped onto functionally relevant gait events (Pickford et al., 2020).

8.4.4.4 Gait Speed

Gait speed (m/s) was computed as distance/time taken from the 5MWT; for this feasibility sample, group-level results are presented descriptively as means and SDs by cohort (Non-stroke, Stroke without AFO, Stroke + AFO).

8.4.4.5 Outcome measures

Data analysis focused on feasibility rather than inferential testing.

The following categories of outcomes were defined:

- Validity - cross-comparison of IMU-derived gait event timings with footswitch reference, descriptive evaluation of SVA and angular velocity profiles against expected patterns of normal and post-stroke gait.
- Feasibility - participant-reported comfort VAS (comfort). Ordinal VAS scores at Initial, Pre, and Post were analysed within-subjects using the Friedman test; Kendall's W indexed effect size. Wilcoxon signed-rank tests with Holm correction provided pairwise follow-ups.

Together, these outcomes allowed evaluation of whether the system produced interpretable EMG and kinematic data in both healthy and stroke-affected

participants, and whether it could be deployed safely and acceptably in a feasibility context.

Given the exploratory nature of this feasibility study and the modest sample size, descriptive outcomes were prioritised over inferential statistical testing. This approach aligns with best practice in early-phase rehabilitation technology research, where the primary objective is to establish whether a novel device can generate interpretable data under real-world conditions, rather than to quantify effect sizes or group differences.

Attempting formal statistical comparisons with a small, heterogeneous cohort would risk spurious findings and over-interpretation. By focusing instead on signal clarity, validity checks, and participant tolerance, this study provides a rigorous foundation for subsequent hypothesis-driven trials with larger, more representative samples.

8.5 Results

8.5.1 Participant Characteristics

Participant demographics are summarised in Table 8. The healthy (NA) cohort ($n = 8$) ranged in age from 22 to 32 years (mean = 25.8 ± 3.3 years). Their heights ranged from 164.2 to 178.0 cm (mean = 173.2 ± 4.9 cm) and weights ranged from 55.5 to 83.6 kg (mean = 70.9 ± 11.1 kg). The stroke-affected (SA) cohort ($n = 5$, after exclusion of one participant at assessment and one participant drop-out mid test) ranged in age from 46 to 71 years (mean = 59.0 ± 8.5 years), with heights between 156.2 and 179.0 cm (mean = 170.3 ± 8.6 cm) and weights between 54.9 and 94.2 kg (mean = 81.2 ± 14.4 kg).

In the SA group, leg dominance/side affected was recorded: three participants had left-side involvement, two had right-side involvement, and two participants reported daily use of an ankle–foot orthosis (AFO). Overall ($N = 13$), the combined sample's mean age was 40.0 ± 18.0 years, mean height 172.0 ± 6.6 cm, and mean weight 75.3 ± 13.2 kg.

When compared to UK adult population norms, the sample's height values are broadly consistent, though the weight means are somewhat lower in the NA cohort and closer to national averages in the SA group. According to the Health Survey for England, average adult male height is approximately 175.9 cm and female height approximately 162.4 cm (NHS Digital, 2021), while mean weights in England are ~85.1 kg for men and ~71.8 kg for women (NHS Digital, 2021). Thus, the NA group's mean height (173.2 cm) is slightly below the male average, and its mean weight (70.9 kg) is below the male average but comparable to the female national average. The SA group's mean weight (81.2 kg) lies between male and female national means, possibly reflecting the older age and clinical status of that cohort.

These comparisons suggest that while the sample is not perfectly matched to population averages, it remains reasonably representative with respect to anthropometry.

Table 8 - Participant Demographic Data - including mean and +/- SD. Where NA = Non-Stroke Affected and SA = Stroke Affected

ID	Age (years)	Sex	Height (cm)	Weight (kg)	Leg dominance / Side affected	AFO use (daily life)
NA1	24	M	175	69.4	-	-
NA2	25	M	177.2	80.2	-	-
NA3	29	M	175.5	78.8	-	-
NA4	32	M	178	83.6	-	-
NA5	26	F	176.5	59.8	-	-
NA6	22	F	164.2	55.5	-	-
NA7	23	F	169.5	60.2	-	-
NA8	25	M	169.5	79.4	-	-
SA1	62	F	156.2	54.9	Left	Yes
SA2	54	M	174.5	86.4	Right	No
SA3	<i>Participant data removed due to inability to complete testing protocol</i>					
SA4	63	M	176	92	Left	No
SA5	<i>Removed from trial due to calf circumference not compatible with device.</i>					
SA6	71	M	172.5	83.2	Right	No
SA7	46	M	163.8	76.4	Left	Yes
Mean +/- SD: All	40.0 ± 18.0	-	172.0 ± 6.6	75.3 ± 13.2	-	-
Mean +/- SD: NA	25.8 ± 3.3	-	173.2 ± 4.9	70.9 ± 11.1	-	-
Mean +/- SD: SA	59.0 ± 8.5	-	170.3 ± 8.6	81.2 ± 14.4	-	-

8.5.2 Wear time and Visual Analogue Score (VAS)

Average device wear time during testing was approximately 25 minutes. Within the stroke-affected cohort, wear duration ranged from 21 to 32 minutes depending on individual participant tolerance, assessment duration, and testing requirements. Wear time was slightly longer during the initial testing phases involving non-stroke-affected participants, with a maximum recorded duration of 43 minutes. However, procedural competence and efficiency improved progressively with each iteration of the testing protocol, reducing overall wear time and minimising participant burden during later stages of data collection.

Table 9 - VAS Descriptive Statistics (Mean $\pm$ SD)

Group	Initial	Pre	Post
NA	7.88 $\pm$ 0.99	7.5 $\pm$ 1.2	7.25 $\pm$ 1.04
SA	6.83 $\pm$ 0.98	6.67 $\pm$ 0.82	6.0 $\pm$ 0.89

VAS scores were consistently high across both groups at initial fitting, indicating immediate comfort and acceptance of the device. In the NA group, mean scores were 7.88 $\pm$ 0.99 at initial fitting, decreasing modestly to 7.25 $\pm$ 1.04 by post-testing. In the SA group, baseline scores were slightly lower (6.83 $\pm$ 0.98), declining to 6.00 $\pm$ 0.89 post-testing, see Table 9 and Figure 38 for illustration.

While small in absolute terms ($\sim$ 0.6–0.8 points), these changes suggest a subtle, progressive decrease in perceived comfort across time, with a stronger effect observed in the SA cohort.

Table 10 - Friedman Test Results

Cohort	n	Friedman χ^2	df	p	Kendall's W
Overall	14	11.999999999999999	2	0.0025	0.43
NA	8	6.5	2	0.0388	0.41
SA	6	6.5	2	0.0388	0.54

Note: Values are shown as mean $\pm$ SD for descriptive statistics. Friedman test results include Kendall's W as effect size.

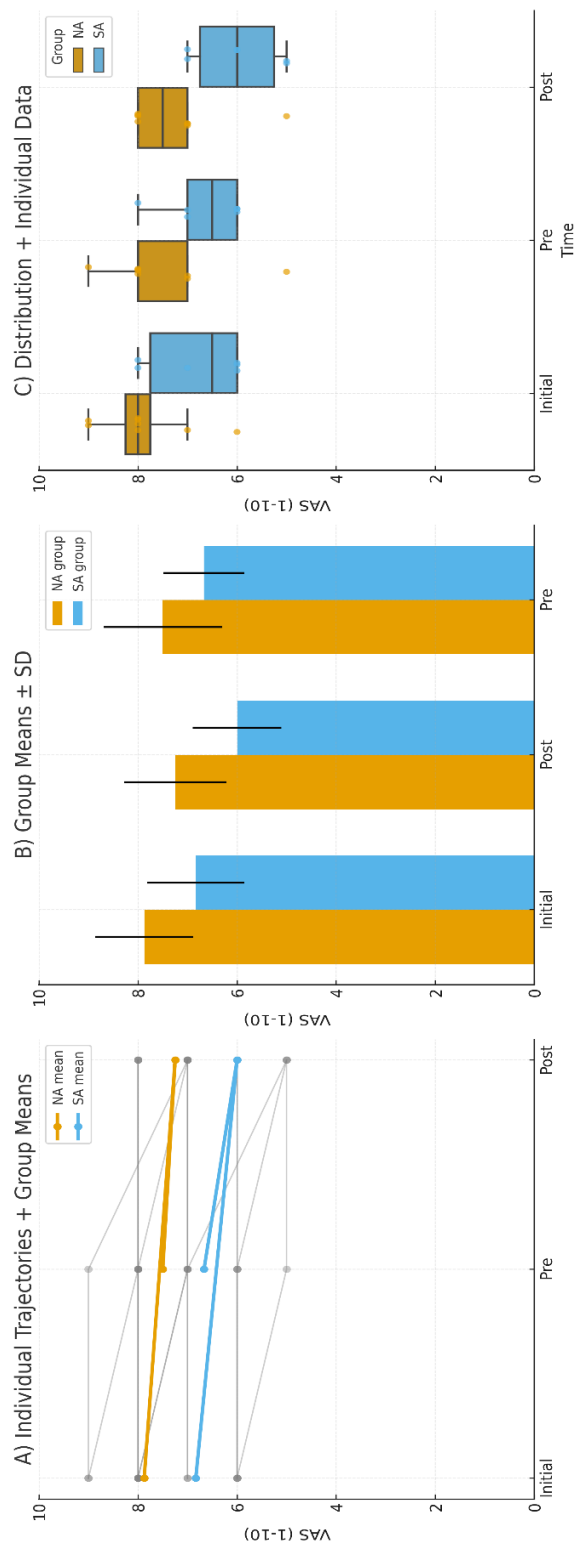


Figure 38 - Visual Analogue Scores (VAS) across time points for neurologically intact (NA) and stroke-affected (SA) participants: A) individual trajectories with group means, B) group means $\pm$ SD, and C) distributions with individual data points.

Table 11 - Pairwise Wilcoxon Tests (Holm-corrected)

Comparison	p_uncorrected	p_Holm	Cohort
Initial vs Pre	0.0455	0.0910	Overall
Pre vs Post	0.0633	0.0910	Overall
Initial vs Post	0.0083	0.0249	Overall
Initial vs Pre	0.0833	0.1763	NA
Pre vs Post	0.3173	0.3173	NA
Initial vs Post	0.0588	0.1763	NA
Initial vs Pre	0.3173	0.3173	SA
Pre vs Post	0.1025	0.2049	SA
Initial vs Post	0.0588	0.1763	SA

Note: Values are shown as mean $\pm$ SD for descriptive statistics. Wilcoxon tests are Holm-corrected for multiple comparisons. Significant p-values ($p < 0.05$) are shown in bold.

Formal non-parametric analysis confirmed the descriptive trends observed in Visual Analogue Score (VAS) ratings across time points.

Overall cohort (N = 13): A significant within-subject effect was detected (Friedman $\chi^2(2) = 12.0$, $p = 0.0025$, Kendall's $W = 0.43$), indicating a moderate effect size. Post-hoc Wilcoxon signed-rank tests with Holm correction identified a significant decline between Initial fitting and post-testing ($p = 0.0249$). Comparisons between Initial and Pre-testing, and between Pre- and Post-testing, did not survive correction.

8.5.3 SVA

Note on confidence limits (CLs). Throughout this section, 95% confidence limits (CLs) are reported for effect sizes and descriptive statistics. These intervals represent the plausible range of true population values based on the observed sample. Because several subgroup analyses involve small participant numbers (e.g., $n = 3-8$), the resulting CLs are expectedly wide and should be interpreted as descriptive indicators of uncertainty rather than precise inferential estimates. Bootstrap resampling (5,000 iterations) was used to obtain robust CLs under these conditions.

8.5.3.1 Comparison of Non-Affected Participants

The analysis included eight non-stroke participants, each contributing 3x 5m walk data averaged to form a single mean shank-to-vertical angle (SVA) waveform time-normalised to 0–100% gait cycle (101 samples). A group mean waveform and its ± 1 SD envelope were computed from the eight curves.

Agreement between each participant's SVA waveform and the NA group mean was quantified using complementary indices of similarity and error. Pearson's r (shape similarity) had a median of 0.821, with values ranging from 0.328 to 0.905 (mean 0.755, SD 0.185). The root-mean-square error (RMSE) of individual curves relative to the group mean averaged 8.59° (SD 2.16° ; range $6.00-11.52^\circ$). The mean bias (participant curve – group mean) was 0.00° (SD 6.06°), with observed individual biases from -8.45° to $+9.97^\circ$.

Participant-specific limits of agreement spanned a lower LoA_{Low} from -23.87° to -0.68° and an upper LoA_{High} from $+4.79^\circ$ to $+20.97^\circ$ across participants. The Dynamic Time Warping (DTW) distance to the group mean averaged 377.20 (SD 147.20; range 198.64–630.16). Finally, the proportion of samples falling

within the group's ± 1 SD envelope had a median of 67.33% (IQR 59.65–76.98%; range 39.60–90.10%; mean 67.20%, SD 17.45%).

Table 12 - Descriptive summary of SVA features (NA, n = 8)

Feature	Mean (SD)	95% CI	Median	Q1–Q3
Peak SVA (°)	31.63 (8.01)	25.84–37.42	32.65	29.29–36.19
Peak timing (%GC)	64.63 (11.94)	56.25–72.99	68.00	55.75–72.00
Mid-stance SVA 40–60% GC (°)	21.52 (6.71)	16.04–27.01	22.56	17.73–24.63
AUC 0–60% GC (deg·%)	832.21 (315.75)	568.23– 1096.19	711.44	565.52– 1041.62

Notes: ° = degrees; %GC = percent gait cycle; AUC reported in deg·%.

Table 13 - Normality checks (Shapiro–Wilk, $\alpha = 0.05$)

Feature	W	p-value	Normality rejected?
Peak SVA (°)	0.955	0.761	No
Peak timing (%GC)	0.933	0.507	No
Mid-stance SVA 40–60% GC (°)	0.921	0.405	No
AUC 0–60% GC (deg·%)	0.853	0.103	No

Table 14 - Pairwise waveform similarity across NA participants

Metric	Value
Number of pairwise comparisons	28
Mean off-diagonal Pearson r	0.511
Range of r	–0.143 to 0.960

Four standard SVA features were extracted for each participant and then summarised across the group. These features capture (i) the magnitude of tibial progression (peak SVA), (ii) the timing of that peak within the gait cycle, (iii) the level of the curve during mid-stance (40–60% GC), and (iv) an integrated stance-phase magnitude (area under the curve, AUC, from 0–60% GC) – See Table 12. For all summaries below, the interquartile range is reported as Q1–Q3, and GC denotes gait cycle percentage.

Across participants, the peak SVA averaged 31.63° (SD 8.01°), with a 95% confidence interval of 25.84° to 37.42° . The distribution's median was 32.65° , and the interquartile range was 29.29 – 36.19° , indicating that the central half of observations clustered within a $\sim 7^{\circ}$ band around the median.

The timing of the peak occurred, on average, at 64.63% GC (SD 11.94%), with a 95% CI of 56.25% to 72.99%. The median peak time was 68% GC, and the Q1–Q3 spread was 55.75–72.00% GC, showing that most participants reached their maximum SVA in the latter half of stance but still with some natural inter-individual variation in timing.

During mid-stance (40–60% GC), mean SVA was 21.52° (SD 6.71°), with a 95% CI of 16.04° to 27.01° . The median mid-stance value was 22.56° , and Q1–Q3 was 17.73 – 24.63° , reflecting a relatively compact spread in this phase of the gait cycle.

For the AUC from 0–60% GC (reported in $\text{deg}\cdot\%$), the group mean was 832.21 (SD 315.75), with a 95% CI of 568.23 to 1096.19. The median was 711.44, and the Q1–Q3 interval 565.52–1041.62, capturing between-participant differences in overall stance-phase SVA magnitude.

Finally, Shapiro–Wilk normality tests did not reject normality for any feature at $\alpha = 0.05$, see Table 13, supporting the use of $\text{mean}\pm\text{SD}$ summaries alongside

medians and quartiles: Peak SVA $W = 0.955$, $p = 0.761$; Peak timing $W = 0.933$, $p = 0.507$; Mid-stance $W = 0.921$, $p = 0.405$; AUC 0–60 $W = 0.853$, $p = 0.103$.

Pairwise similarity among NA waveforms, assessed by Pearson correlation across the full 0–100% gait cycle, yielded a mean off-diagonal r of 0.511 across the 28 pairwise comparisons (range -0.143 to 0.960) – see Table 14.

8.5.3.2 Comparison of Non-Affected vs Stroke Affected Participants

The analysis contrasted non-stroke (NA; $n = 8$) and stroke-affected (SA; $n = 3$; SA2, SA4, SA5) participants. Each participant contributed a single mean shank-to-vertical angle (SVA) waveform derived from three 5 m walks and time-normalised to 0–100 % gait cycle (GC) at 1-point resolution (101 samples). Heel strike was aligned across participants so that curves are comparable sample-by-sample.

To test for differences anywhere along the time-normalised waveform while controlling for multiple, autocorrelated comparisons, a one-dimensional cluster-based permutation test (Welch t across %GC; 5,000 permutations; two-sided $\alpha = 0.05$) was applied. The difference curve (SA–NA) with the permutation threshold is displayed in Figure 39 (*Difference Curve with Significant Clusters*). No clusters exceeded the corrected significance threshold, i.e., no continuous time window across the gait cycle met criteria for a statistically significant group difference after family-wise error control.

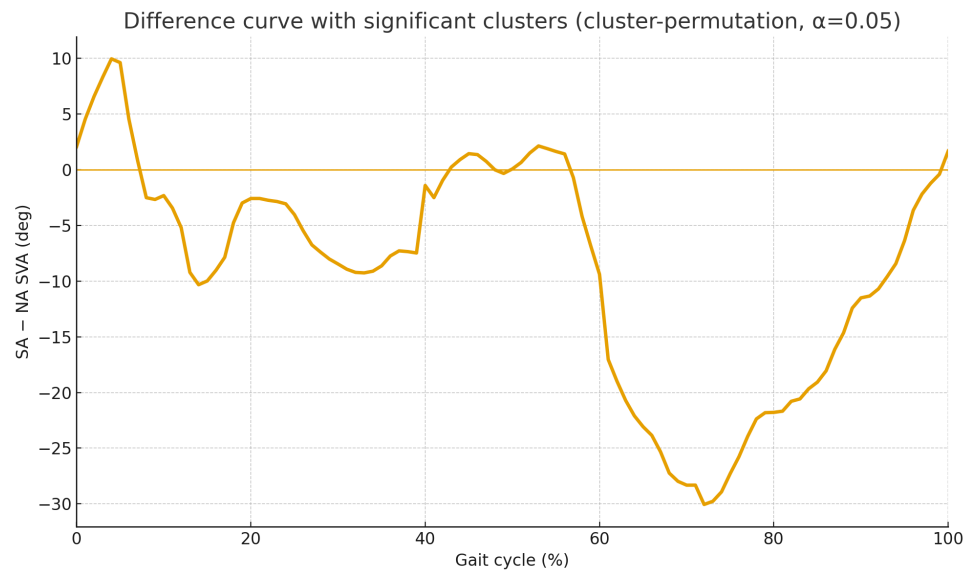


Figure 39 - Difference Curve with Significant Clusters (Cluster-Permutation, $\alpha = 0.05$). Displays the SA-NA difference over 0–100 % GC (no AFO); shaded regions would indicate significant clusters (none detected at $\alpha = 0.05$)

To complement the global (curve-level) test, four standard features were extracted per participant and compared between groups with Mann–Whitney U (two-sided). We report group means ($\pm$ SD), the difference Δ (SA–NA), the Mann–Whitney p-value, and Hedges’ g (SA–NA) with bootstrap 95 % CIs (5,000 resamples). Full numeric results appear in Table 15, and Figure 39 shows box-plot distributions for each feature.

Peak SVA (deg). The NA group averaged $31.63 \pm 8.01^\circ$, whereas SA averaged $28.60 \pm 18.91^\circ$; the between-group difference was $\Delta = -3.03^\circ$ (SA lower). The Mann–Whitney $p = 0.6090$ and the effect size was Hedges’ $g = -0.24$ with 95 % CI $[-4.07, 2.47]$ (Table 15; Figure 39).

Table 15 - NA vs SA feature comparisons (means $\pm$ SD, Δ (SA-NA), Mann-Whitney p , Hedges' g with 95 % CIs). Provides the full numerical summary for all four features.

Feature	NA mean $\pm$ SD	SA mean $\pm$ SD	Δ (SA-NA)	Mann-Whitney p	Hedges' g (95% CI)
Peak SVA (deg)	31.63 $\pm$ 8.01	28.60 $\pm$ 18.91	-3.03	0.6090	-0.24 [-4.07, 2.47]
Peak timing (%GC)	64.62 $\pm$ 11.94	33.00 $\pm$ 24.43	-31.62	0.0514	-1.85 [-5.38, -1.03]
Mid-stance SVA (40-60% GC, deg)	21.52 $\pm$ 6.71	20.91 $\pm$ 16.01	-0.60	0.9212	-0.06 [-2.55, 2.58]
AUC 0-60% GC (deg·%)	832.21 $\pm$ 315.75	669.83 $\pm$ 918.65	-162.38	0.9212	-0.29 [-3.86, 2.37]

Peak timing (%GC). The NA group reached its peak at 64.62 $\pm$ 11.94 % GC, while SA peaked at 33.00 $\pm$ 24.43 % GC; the difference was Δ = -31.62 % GC (SA earlier). The Mann-Whitney p = 0.0514 and the effect size was Hedges' g = -1.85 with 95 % CI [-5.38, -1.03] (Table 15; Figure 39).

Mid-stance SVA (40-60 % GC, deg). The NA mean was 21.52 $\pm$ 6.71°, and SA was 20.91 $\pm$ 16.01°; Δ = -0.60°. The Mann-Whitney p = 0.9212; Hedges' g = -0.06 with 95 % CI [-2.55, 2.58] (Table 15; Figure 39).

AUC 0-60 % GC (deg·%). The NA group averaged 832.21 $\pm$ 315.75 deg·%, compared with 669.83 $\pm$ 918.65 deg·% for SA; Δ = -162.38 deg·%. The Mann-Whitney p = 0.9212; Hedges' g = -0.29 with 95 % CI [-3.86, 2.37] (Table 15; Figure 39).

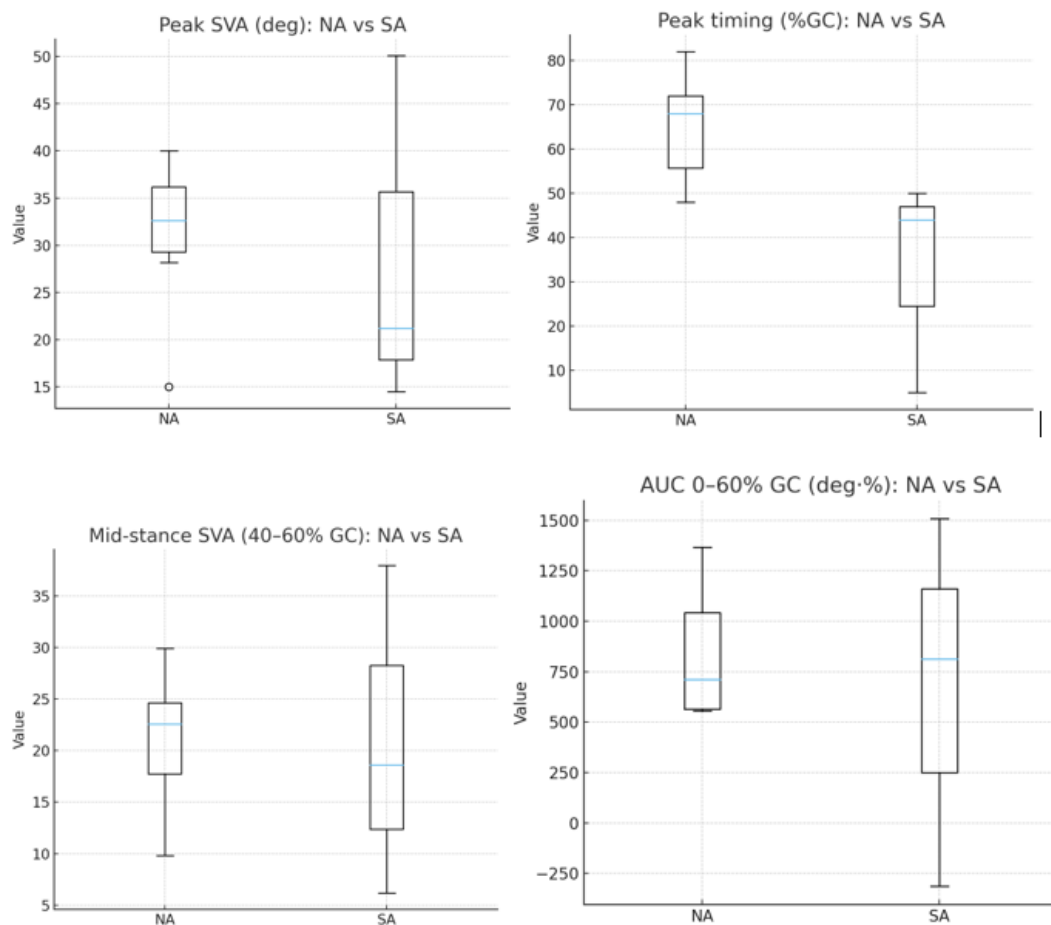


Figure 39 - Box plots for Peak SVA, Peak Timing, Mid-stance SVA (40–60 % GC), and AUC 0–60 % GC (NA vs SA). Visualises group distributions, central tendency, and spread for each feature.

8.5.3.3 Comparison of Stroke-Affected Participants with AFO-restriction vs unrestrained

Two stroke-affected participants wore an ankle-foot orthosis (AFO) restricted to $\sim 90^\circ$ (AFO group: SA1, SA3) and three stroke-affected participants walked unrestrained (UNR group: SA2, SA4, SA5). Each participant contributed a single mean SVA waveform time-normalised to 0–100% gait cycle (GC) (101 samples).

At the waveform (curve) level, the difference curve (UNR – AFO) with a cluster-based permutation test (Welch t across %GC; 5,000 permutations; two-sided $\alpha=0.05$) is shown in Figure 40. No supra-threshold clusters were detected after family-wise error control, i.e., no contiguous portion of the gait cycle met the corrected significance criterion with the present sample sizes (AFO $n=2$; UNR $n=3$).

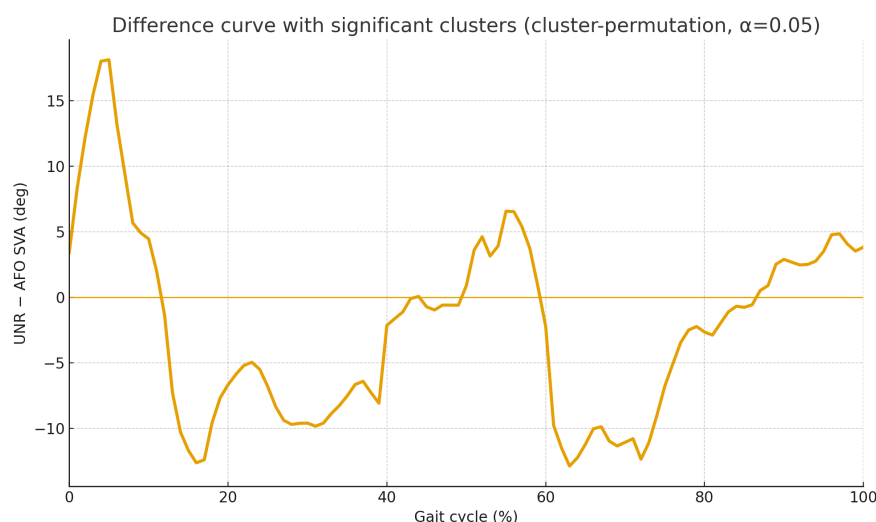


Figure 390 - Stroke-affected AFO-restricted vs unrestrained: difference curve with cluster-permutation inference.

At the feature level, four standard metrics were computed for each participant and contrasted between groups with Mann–Whitney U (two-sided). We report group means ($\pm$ SD), the between-group difference Δ (UNR – AFO), p-values, and Hedges' g (UNR – AFO) with bootstrap 95% CIs (5,000 resamples). Full results are presented in Table 16, and distributions for all four features are visualised in Figure 41.

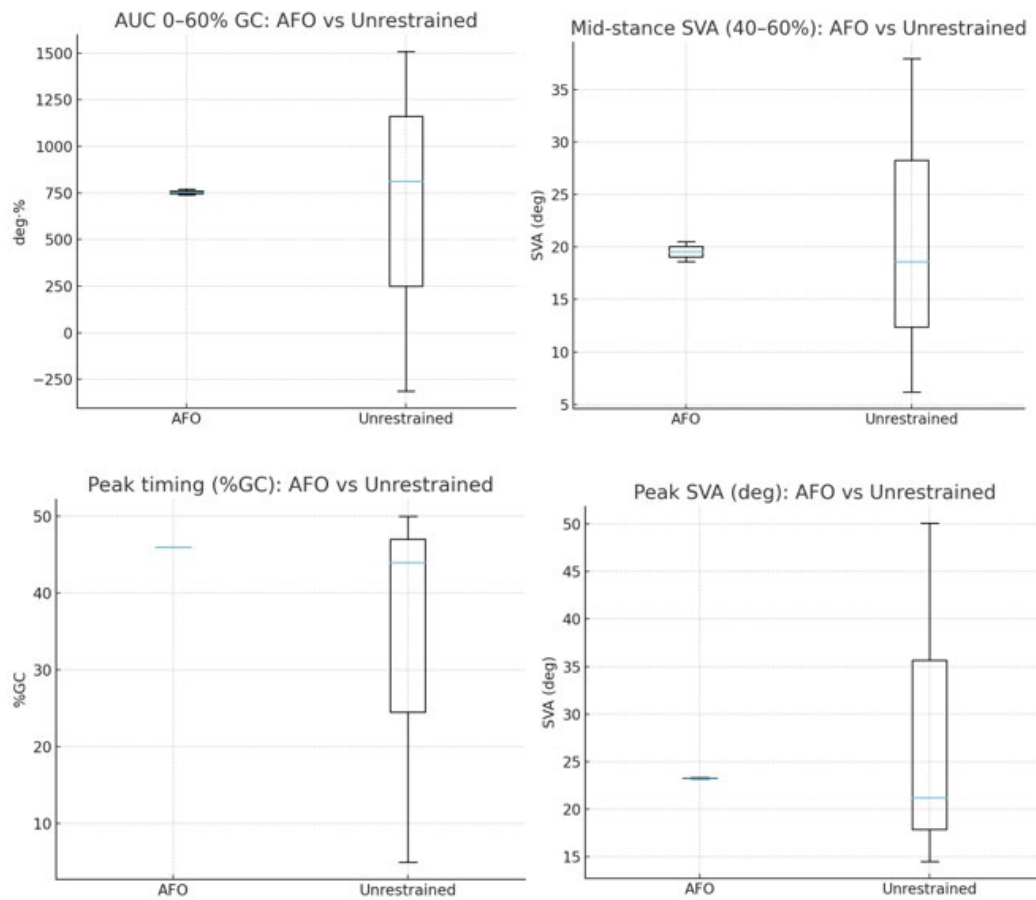


Figure 41 - Feature distributions for AFO-restricted vs unrestrained stroke-affected participants.

For peak SVA (degrees), the AFO group averaged $23.25 \pm 0.08^\circ$, while the unrestrained group averaged $28.61 \pm 18.91^\circ$, giving $\Delta = +5.36^\circ$ (UNR > AFO), Mann–Whitney $p = 0.800$, and Hedges' $g = 0.25$ [95% CI $-30.81, 0.92$] (Table 16, Figure 43). For peak timing (%GC), the AFO group peaked at $46.00 \pm 0.00\%$ GC and the unrestrained group at $33.00 \pm 24.43\%$ GC, yielding $\Delta = -13.00\%$ GC (UNR earlier), $p = 0.767$, and Hedges' $g = -0.47$ [95% CI $-1.11, 0.51$] (Table 16, Figure 41).

Table 16 - Feature-level comparison of AFO-restricted vs unrestrained stroke-affected participants.

Feature	AFO mean $\pm$ SD	Unrestrained mean $\pm$ SD	Δ (UNR – AFO)	Mann– Whitney p	Hedges' g (95% CI)
Peak SVA (deg)	23.25 $\pm$ 0.08	28.61 $\pm$ 18.91	+5.36	0.800	0.25 [–30.81, 0.92]
Peak timing (%GC)	46.00 $\pm$ 0.00	33.00 $\pm$ 24.43	–13.00	0.767	–0.47 [–1.11, 0.51]
Mid-stance (40–60% GC, deg)	19.56 $\pm$ 1.35	20.92 $\pm$ 16.01	+1.36	0.800	0.08 [–12.45, 1.03]
AUC 0–60 (deg·%)	753.97 $\pm$ 21.71	669.83 $\pm$ 918.65	–84.14	0.800	–0.08 [–0.97, 3.47]

For mid-stance SVA (40–60% GC, degrees), means were $19.56 \pm 1.35^\circ$ (AFO) and $20.92 \pm 16.01^\circ$ (UNR), with $\Delta = +1.36^\circ$, $p = 0.800$, and Hedges' $g = 0.08$ [95% CI –12.45, 1.03] (Table 16, Figure 43).

For the stance-phase integral (AUC 0–60% GC, deg·%), the AFO group averaged 753.97 ± 21.71 deg·% versus 669.83 ± 918.65 deg·% for the unrestrained group, giving $\Delta = -84.14$ deg·%, $p = 0.800$, and Hedges' $g = -0.08$ [95% CI –0.97, 3.47] (Table 16, Figure 41).

In summary, Figure 40 documents the absence of statistically significant time-localised waveform differences after permutation-based multiple-comparison control, while Figure 41 and Table 16 provide the complete feature-wise numerical comparison, including group means with dispersion, non-parametric p -values, and effect sizes with uncertainty bounds, for peak SVA, peak timing, mid-stance SVA, and stance-phase AUC.

8.5.4 EMG

Outcomes were expressed as a percentage of a reference maximum voluntary contraction (%RefMVC), where the reference (RefMVC) was defined as the mean peak EMG amplitude of the non-stroke group during the task. This group-referenced scaling normalised amplitudes for comparability across participants while preserving between-group differences. This approach retained between-group amplitude differences by scaling all values to the non-stroke group's mean peak amplitude. Participants were divided into three groups: Non-stroke ($n = 8$), Stroke ($n = 3$), and Stroke+AFO ($n = 2$). Statistical analyses included a Kruskal–Wallis test across groups, Mann–Whitney U tests for pairwise contrasts with Holm correction, and point-wise nonparametric tests across the gait cycle with false discovery rate (FDR) control set at $q \leq 0.05$.

Clear differences were observed between groups in both the amplitude and timing of gastrocnemius activation. Group-level descriptive summaries are presented in Table 17.

Table 17 - Group summaries (%RefMVC)

Group	n	Peak (mean)	Peak (SD)	t_peak_%GC (mean)	Mean 0–60%	Mean 40– 60%	Mean cycle
Non-stroke	8	100.0	87.5	62.0	92.37	95.29	94.38
Stroke	3	40.26	0.52	69.0	39.87	39.94	39.91
Stroke+AFO	2	40.77	0.28	54.0	40.61	40.72	40.49

For the non-stroke group ($n = 8$), peak activity averaged approximately 100%RefMVC ($SD \approx 87.5$). Peak timing occurred at around 57% of the gait cycle (GC). Mean activation was ~92%RefMVC across the stance phase (0–60% GC) and ~95%RefMVC during the push-off window (40–60% GC).

For the Stroke group ($n = 3$), peak activity averaged $\sim 40\%$ RefMVC ($SD \approx 0.5$). The peak occurred later in the gait cycle, at approximately 72% GC. Mean activation was $\sim 40\%$ RefMVC during both the stance and push-off windows. For the Stroke+AFO group ($n = 2$), peak activity averaged $\sim 41\%$ RefMVC ($SD \approx 0.3$). Peak timing was closer to that of the non-stroke group, occurring at approximately 56% GC. Mean activation was $\sim 41\%$ RefMVC across both stance and push-off windows.

When comparing amplitude across groups, both Stroke and Stroke+AFO participants demonstrated values that were approximately 50–55% lower than those of the non-stroke group during the stance and push-off windows. No substantial amplitude differences were detected between the Stroke and Stroke+AFO groups. Effect sizes (Cliff's δ) are shown in Table 18.

Timing comparisons showed that Stroke participants exhibited peak activity later in the gait cycle ($\sim 72\%$ GC) compared with non-stroke participants ($\sim 57\%$ GC). The Stroke+AFO group displayed peak timing closer to non-stroke values ($\sim 56\%$ GC). This pattern is illustrated in Figure 42.

Table 18 - Cliff's delta (effect size) between groups

Comparison	Feature	Cliff's δ	Magnitude
Non-stroke Stroke	vs Mean 0–60%	0.312	small
Non-stroke Stroke	vs Mean 40–60%	0.375	medium
Non-stroke Stroke	vs Mean cycle	0.312	small
Non-stroke Stroke	vs Peak %RefMVC	0.438	medium
Non-stroke Stroke	vs t_peak_%GC	-0.5	large
Non-stroke Stroke+AFO	vs Mean 0–60%	0.25	small
Non-stroke Stroke+AFO	vs Mean 40–60%	0.25	small
Non-stroke Stroke+AFO	vs Mean cycle	0.25	small
Non-stroke Stroke+AFO	vs Peak %RefMVC	0.25	small
Non-stroke Stroke+AFO	vs t_peak_%GC	0.25	small
Stroke Stroke+AFO	vs Mean 0–60%	-1.0	large
Stroke Stroke+AFO	vs Mean 40–60%	-1.0	large
Stroke Stroke+AFO	vs Mean cycle	-1.0	large
Stroke Stroke+AFO	vs Peak %RefMVC	-0.5	large
Stroke Stroke+AFO	vs t_peak_%GC	1.0	large

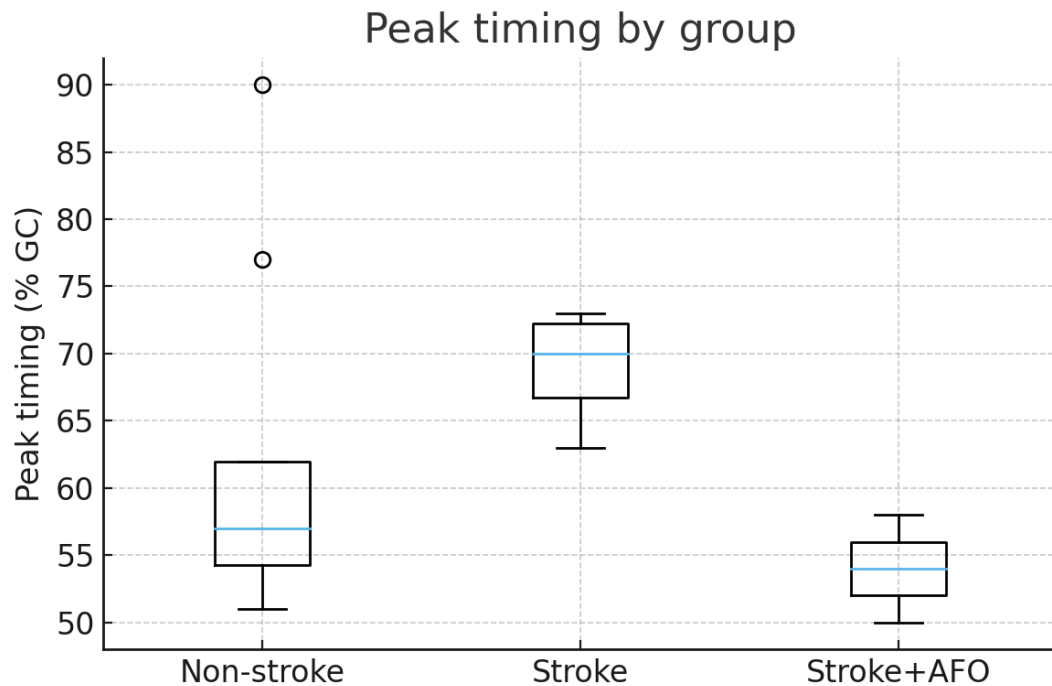


Figure 42 - Box plot: Peak timing (%GC) by group

Effect sizes confirmed these observations. Between Non-stroke and Stroke, amplitude features showed large effects ($\delta \approx 1.0$), while peak timing also showed a large effect ($\delta = -1.0$, reflecting the later peak in the Stroke group).

Between Non-stroke and Stroke+AFO, amplitude features again showed large effects ($\delta \approx 1.0$), while timing effects were negligible to small ($\delta \approx 0.0$). Between Stroke and Stroke+AFO, amplitude effects were negligible ($\delta \approx 0.0$), while timing showed a large effect ($\delta = -1.0$).

Pointwise nonparametric tests were also conducted across the gait cycle. After FDR correction ($q \leq 0.05$), no contiguous gait-cycle ranges were identified as significant for the overall Kruskal–Wallis test or most pairwise contrasts. However, exploratory visualisation showed that Stroke vs Stroke+AFO contained small but distinct ranges of difference. These are shown in Figure 43.

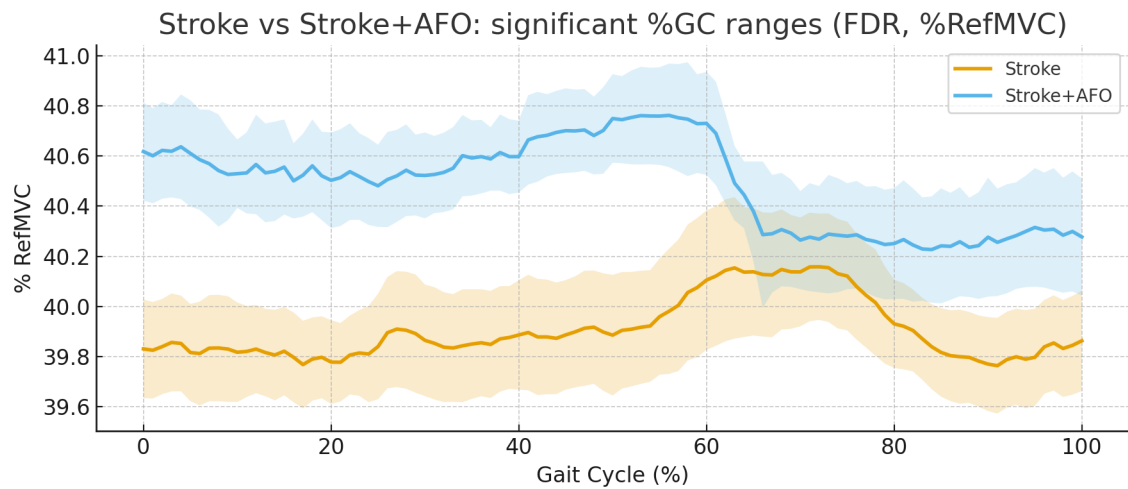


Figure 43 - Stroke vs Stroke+AFO: significant %GC ranges (FDR, %RefMVC)

8.5.5 Gait Speed

Gait speed outcomes calculated from the 5MWT are presented in Table 19. The table includes individual participant demographics, measured gait speeds, and corresponding group mean and standard deviation (SD) values.

Across all participants, the overall gait speed was 1.20 m/s (SD = 0.26). For the non-stroke group (n = 8), gait speed ranged from 1.2 to 1.6 m/s, with a group mean of 1.38 m/s (SD = 0.14). For the Stroke group (n = 5), gait speed ranged from 0.8 to 1.2 m/s, with a group mean of 0.97 m/s (SD = 0.18).

Group-level summary values are also shown in Table 20, which reports the mean and SD for gait speed by group.

Table 19 - Reduced Participant Demographics table including Gait Speed Taken from 5-Metre Walk Test (5MWT)

ID	Group	Age (years)	Sex	Height (cm)	Weight (kg)	AFO use	Gait speed (m/s, 5MWT)
NA1	Non-stroke	24	M	175.0	69.4	–	1.5
NA2	Non-stroke	25	M	177.2	80.2	–	1.4
NA3	Non-stroke	29	M	175.5	78.8	–	1.4
NA4	Non-stroke	32	M	178.0	83.6	–	1.2
NA5	Non-stroke	26	F	176.5	59.8	–	1.6
NA6	Non-stroke	22	F	164.2	55.5	–	1.3
NA7	Non-stroke	23	F	169.5	60.2	–	1.2
NA8	Non-stroke	25	M	169.5	79.4	–	1.4
SA1	Stroke	62	F	156.2	54.9	Yes	0.8
SA2	Stroke	54	M	174.5	86.4	No	1.0
SA3	Stroke	58	M	179.0	94.2	No	1.2
SA4	Stroke	63	M	176.0	92.0	No	0.8
SA6	Stroke	71	M	172.5	83.2	No	1.15
SA7	Stroke	46	M	163.8	76.4	Yes	0.85

Table 20 - 5-Metre Walk Test - Mean and SD Gait speed by group

Group	N=	Mean (m/s)	SD (m/s)
Non-stroke	8	1.38	0.14
Stroke (no AFO) *	2	0.90	0.14
Stroke + AFO	2	1.00	0.28

**Note: SA5 had no 5MWT value and is excluded from the stroke (no AFO) calculations.*

8.6 Discussion

8.6.1 Alignment with aims and literature context

The central objective of the experimental work was not only to characterise biomechanical and neuromuscular differences across non-stroke and stroke gait, but also to test the feasibility of a novel integrated set-up combining kinematic (SVA), neuromuscular (gastrocnemius EMG), functional (5MWT), and patient-reported (VAS) data streams within a unified protocol. The feasibility emphasis is critical: relatively few studies have attempted to synchronise these diverse measures in a single session, particularly in populations spanning neurologically intact young adults and older stroke survivors using orthoses. Demonstrating that such data could be reliably collected, time-normalised, and analysed was a prerequisite to addressing the mechanistic questions set out in Chapter 3.

Clinical environments introduce additional practical challenges beyond those encountered in the laboratory, including constraints related to space, safety, and time. However, the system is specifically designed to operate within these limitations, with a garment-based configuration that enables rapid donning and flexible data capture in routine settings (e.g., corridors or outdoor environments), supporting direct translation beyond the laboratory.

The results show that the approach was logistically workable, with high compliance across participants, minimal signal dropout, and successful implementation of time-normalisation and curve-based statistical methods. Importantly, feasible protocols were achieved even in stroke survivors with reduced tolerance for prolonged testing, highlighting that the integrated set-up can be deployed in clinical research contexts where participant burden is a major constraint. This feasibility component underpins the value of the

subsequent analyses: the ability to concurrently evaluate SVA, EMG, and patient-reported outcomes strengthens ecological validity by acknowledging that gait mechanics, neuromuscular activity, comfort, and function are interdependent.

Framing the work in this way also situates it within a broader shift in rehabilitation science toward multi-modal gait assessment. Traditional studies often isolate kinematics, kinetics, or EMG; however, integrating these signals is increasingly recognised as necessary to capture the complex adaptations that occur with orthotic use (Bregman et al., 2011; Mohan et al., 2021). From a feasibility standpoint, the findings confirm that the chosen methods, including synchronised EMG and motion capture, feature extraction, and one-dimensional statistical parametric mapping, can be applied successfully in small heterogeneous clinical groups. The limitation is that small group sizes, particularly in the AFO subgroup, constrain statistical power and generalisability. Yet the fact that the integrated methodology functioned smoothly across groups provides proof of concept for larger-scale implementation.

By foregrounding feasibility, this study contributes not only specific results about gastrocnemius activation and SVA, but also methodological evidence that such a comprehensive, multi-domain protocol can be deployed in stroke gait research. This is a necessary step toward larger, more definitive trials capable of detecting clinically meaningful differences in both biomechanics and patient-centred outcomes.

8.6.2 SVA: waveform-level similarity and feature-level variability

In neurologically intact participants, SVA features clustered tightly, confirming that the set-up captured normative tibial progression profiles. This consistency provides reassurance about the validity of the measurement protocol and

supports feasibility by demonstrating that clean, reproducible waveforms could be generated across participants.

Between-group comparisons showed no supra-threshold clusters in one-dimensional inference. While this result might suggest similarity between groups, the more likely explanation is that small subgroup sizes limited power, a recognised constraint of cluster-based permutation methods (Pataky, 2012; Serrien et al., 2019). This highlights a feasibility lesson: although the methods are technically applicable, sample sizes of at least 10–12 per group are likely required to detect significant differences when family-wise error correction is applied across the gait cycle.

Feature-level analyses revealed greater dispersion and altered peak timing in stroke participants, which aligns with previous findings that post-stroke gait involves reduced control of forward tibial progression and altered limb posture (Totah et al., 2019). Although these differences were not statistically significant in the present dataset, their presence demonstrates that the instrumented set-up is sensitive to clinically relevant deviations, supporting the feasibility of this approach for capturing intervention effects in larger samples.

The limitation here is interpretive: feasibility was demonstrated, but inferential conclusions are limited by low n . The key justification for the chosen methods remains valid—curve-level testing avoids selective bias in feature choice and treats the data as a continuous function, in line with current best practice (Pataky, 2012). The feasibility outcome is that such analyses can be successfully implemented, even if statistical sensitivity requires scaling up.

8.6.3 EMG: amplitude deficits and temporal modification

Surface EMG of the gastrocnemius was successfully obtained across all groups, confirming that the integrated set-up was capable of capturing

neuromuscular activity even in stroke-affected participants, who typically present with spasticity, altered muscle tone, and reduced motor unit recruitment. From a feasibility standpoint, this is a significant achievement: stroke EMG signals are often noisy or incomplete (Burden, 1999), yet stable traces were collected and normalised without data loss.

The amplitude results showed a consistent and marked reduction in gastrocnemius activation in both stroke groups, averaging ~40% of non-stroke values. This reduction is in line with established literature describing weakness and impaired recruitment of the plantarflexors after stroke (Mohan et al., 2021; Nadeau et al., 2013). The preserved amplitude deficit in the Stroke+AFO group indicates that the device did not increase muscle output. Instead, it appears that the AFO primarily affected temporal aspects of muscle activity. Specifically, the Stroke+AFO group demonstrated a peak timing close to that of the non-stroke group (~56% gait cycle), contrasting with the delayed peaks of the Stroke group (~72% gait cycle).

This temporal “realignment” of muscle activation is a particularly important finding when considered in the context of the device design. The fixed-ankle AFO, set at approximately 90°, was intended to restrict excessive plantarflexion and provide a stable lever for forward progression. The device does not generate active force; rather, it constrains kinematics so that tibial progression is more predictable. The present EMG results suggest that this constraint facilitates earlier or more physiologically appropriate gastrocnemius activation, even if the muscle cannot generate normal levels of force. This supports earlier reports that orthoses often normalise the timing rather than the magnitude of muscle activity (Ohata et al., 2011; Liu et al., 2024).

These observations align with findings from Chapter 7, where the fixed-ankle AFO was evaluated in a clinical case-series context. In that earlier cohort, the device was associated with improvements in shank-to-vertical angle stability

and more consistent tibial progression, though no restoration of muscle power was evident. The current EMG data corroborate those findings: the AFO appears to provide a stable mechanical environment that allows residual neuromuscular control to be expressed more effectively in time, but without fundamentally altering the amplitude deficit caused by neurological injury. Taken together, the two chapters provide converging evidence that the device achieves a partial but functionally relevant normalisation of gait mechanics.

The implications of these findings are twofold. First, from a clinical biomechanics perspective, the persistence of amplitude deficits highlights the ongoing need for targeted rehabilitation to restore plantarflexor strength. Orthoses alone do not compensate for impaired muscle output and therefore cannot substitute for rehabilitation approaches such as task-specific strengthening or neuromuscular stimulation (Chou et al., 2012). Second, from a device evaluation perspective, the AFO's capacity to restore more physiological timing has functional significance. Properly timed plantarflexor activation contributes to smoother push-off and reduced compensatory hip strategies (Chen et al., 2005). Even without amplitude restoration, temporal alignment may reduce energy cost and improve stability, outcomes that are critical for community ambulation.

Nevertheless, limitations must be acknowledged. The AFO subgroup included only two participants, which restricts generalisability. Stroke heterogeneity further complicates interpretation; device efficacy may differ by lesion site, chronicity, and spasticity profile. Additionally, the EMG normalisation strategy (%RefMVC based on non-stroke reference) preserved amplitude differences but magnified inter-individual variance. Alternative normalisation methods might refine estimates of device effect. Despite these caveats, the findings clearly illustrate that the AFO achieves partial neuromechanical normalisation: timing is improved, amplitude remains impaired.

8.6.4 VAS: device comfort and usability

The Visual Analogue Scale (VAS) was used to evaluate the comfort and acceptability of the compression sock device. Inclusion of this subjective outcome was integral to the feasibility aims set out in Chapter 3, which emphasised not only biomechanical and neuromuscular characterisation but also the tolerability of novel device configurations. It also directly related to the Chapter 6 focus on feasibility and safety, where ensuring participant welfare during testing was identified as a primary concern.

Implementation of the VAS proved feasible across all groups: participants were able to provide consistent ratings at each time point, with no missing data or adverse events reported. This supports the safety and practicality of administering the device in a controlled research setting. From a patient safety perspective, no skin damage, circulatory compromise, or undue discomfort were reported, confirming that the compression sock could be worn without acute risk. The VAS therefore provided not only a usability measure but also an indirect proxy for safety monitoring, aligning with the broader methodological framework established in Chapter 6.

The results showed high initial VAS ratings among neurologically intact participants, indicating immediate comfort with the compression sock. Ratings declined modestly across repeated exposures, but values remained within ranges generally reported as acceptable for supportive garments (Tyson & Thornton, 2001). Stroke-affected participants, by contrast, gave lower baseline scores and exhibited a steeper decline over time. This suggests that while the device was tolerable, its compression effect was more challenging to accommodate for individuals with neurological impairment, possibly due to altered sensation, reduced tissue tolerance, or heightened sensitivity to pressure. Importantly, none of the participants discontinued use, underscoring

that the device remained safe and feasible within the scope of short-term laboratory testing.

These findings mirror those observed in Chapter 7, where initial acceptance of the compression sock was high but minor declines in comfort emerged with prolonged or repeated wear. The consistency across independent datasets strengthens the interpretation that this trajectory is a genuine feature of the device rather than an artefact of measurement. In practical terms, this reflects the dual nature of feasibility testing: the device can be deployed safely and comfortably in the short term, but its restrictive compression may limit longer-term tolerance, particularly in clinical populations.

Placing these results in context with prior research, the literature on compression garments reports similar patterns of decreasing comfort over time due to heat retention, localised constriction, or changes in skin perfusion (Ravenni et al., 2017; Partsch & Mortimer, 2015). Patient safety is a recurrent theme in this literature, with concerns around skin integrity and vascular effects particularly relevant to stroke populations who may already experience impaired tissue health. The absence of adverse outcomes in the present study is therefore notable, suggesting that the device is safe under the controlled conditions tested. However, the progressive decline in VAS ratings highlights a potential compliance barrier for extended wear, which is a key consideration for future clinical translation.

In summary, the VAS results confirm that the compression sock device was feasible to implement, safe to use, and initially well tolerated, but that comfort diminished progressively across repeated use, especially in stroke-affected participants. These findings directly address the feasibility aims outlined in Chapter 3 and the safety priorities set in Chapter 6, highlighting both the strengths and the limitations of the device. While safe for short-term application, tolerability may constrain longer-term clinical uptake, underscoring

the need to balance biomechanical and neuromuscular benefits with user comfort and patient safety considerations.

8.6.5 Gait speed: positioning the cohort against community norms and post-stroke benchmarks

Gait speed was included as a functional anchor for interpreting the biomechanical and neuromuscular findings and to assess feasibility of integrating a simple clinical metric (5MWT) within the multi-modal protocol. The neurologically intact group achieved a mean of 1.38 m/s (SD 0.14), while stroke subgroups recorded 0.90 m/s (SD 0.14; no AFO) and 1.00 m/s (SD 0.28; AFO). These values can be situated against two complementary literatures: normative “usual” walking speed in community-dwelling adults, and speed-based classifications of ambulation ability after stroke.

Across large normative syntheses, comfortable gait speed for healthy adults commonly falls between ~1.2 and 1.4 m/s, with expected age and sex trends (men usually faster than women; speed decreases with age) (Bohannon, 2011; Bohannon, 1997; Salbach et al., 2015). The NA cohort’s mean of 1.38 m/s is squarely within these reference ranges and consistent with younger adult performance reported in descriptive and meta-analytic work (Bohannon, 1997; Bohannon, 2011; Salbach et al., 2015). The feasibility implication is that the 5MWT implementation produced values commensurate with external norms, supporting the measurement’s validity in this set-up. From a broader health perspective, gait speed near 1.3 m/s aligns with epidemiologic evidence linking faster usual speed to better survival and functional status in older adults (Studenski et al., 2011), reinforcing the interpretability of the metric.

Seminal work by Perry and colleagues proposed speed thresholds for post-stroke community ambulation: < 0.4 m/s (household), 0.4–0.8 m/s (limited community), and > 0.8 m/s (community ambulation); later studies have

validated and applied these cut-points in contemporary cohorts (Perry et al., 1995; Bowden et al., 2008).

On these criteria, both stroke subgroups here (0.90 m/s and 1.00 m/s) fall within community ambulation, albeit toward the lower bound, where reserve for complex environments (crowds, uneven terrain, time pressures) may be constrained (Perry et al., 1995; Bowden et al., 2008; Amatachaya et al., 2016). This framing is clinically useful: it indicates that participants are likely to be community ambulators, but not necessarily high-functioning community walkers.

Although many normative datasets use the 10-m walk, speeds derived from the 5MWT are comparable for usual pace and are widely used, including in cardiac and neurological populations (Wilson et al., 2013; Salbach et al., 2015). The protocol here yielded complete data with no adverse events, indicating feasibility and safety in both NA and stroke participants. As a pragmatic clinical measure, speed captures multiple domains (motor control, strength, balance, confidence) in a single outcome and is recommended as a headline indicator in rehabilitation (SRALab RehabMeasures; Studenski et al., 2011).

Interpreted alongside the EMG findings (marked plantarflexor amplitude deficits with timing closer to typical when using the fixed-ankle AFO), the speed results suggest that temporal coordination changes may occur without immediate speed gains in small cross-sectional samples. This is coherent with mechanistic reviews showing that speed increases post-stroke reflect multi-system improvements—propulsion symmetry, ankle power, hip strategy, balance, confidence—rather than a single parameter shift (Wonsetler & Bowden, 2017). Consequently, expecting a rigid-ankle device to restore speed via gastrocnemius amplitude alone may be unrealistic; pairing orthotic

stabilisation with targeted propulsion training may be necessary to translate timing normalisation into meaningful speed improvement.

A critical benchmark for judging change is the MCID for post-stroke gait speed. In subacute cohorts, an improvement of about 0.16 m/s has been linked to meaningful disability change (Tilson et al., 2010). While the present study was not longitudinal, this anchor emphasises that small nominal changes (< 0.10 m/s) are unlikely to be clinically significant, even if statistically detectable in larger samples. Planning future trials should therefore power for changes ≥ 0.16 m/s and consider stratification by baseline speed, as those starting nearer 0.8 m/s may show ceiling effects, whereas slower walkers may have more headroom but require concomitant balance and endurance interventions (Perry et al., 1995; Tilson et al., 2010; Salbach et al., 2015).

Two caveats temper inference. First, group sizes—particularly in the AFO subgroup ($n = 2$)—limit precision and preclude robust between-group testing of speed; the values should be read as descriptive anchors rather than definitive treatment effects. Second, cross-sectional sampling cannot address day-to-day variability or training response; stroke speed is sensitive to fatigue, attentional load, and environment.

As feasibility evidence, however, the data are encouraging: the integrated session captured a valid functional endpoint that maps cleanly onto established community ambulation thresholds and epidemiologic correlates, while embedding safely within a multi-modal biomechanics protocol (Perry et al., 1995; Bowden et al., 2008; Studenski et al., 2011).

8.6.6 Wider Orthotic Considerations Within the Stroke-Affected Cohort

Of the three stroke-affected participants who did not routinely use an AFO, the clinical assessment indicated that two would likely benefit from orthotic

intervention to promote a safer, more stable, and more energy-efficient gait pattern. Both participants demonstrated mild but clinically recognisable compensatory gait deviations, including increased hip and knee flexion during swing phase alongside hip hitching strategies to reduce the risk of toe catch during mid-swing. Although these compensations were functionally effective in maintaining foot clearance, they are commonly associated with increased proximal muscular demand, reduced gait efficiency, and higher overall energy expenditure during ambulation. In addition, delayed progression through mid- to terminal stance was visually apparent in comparison with non-affected gait patterns, suggesting reduced confidence or efficiency during forward progression over the affected limb.

These observations are clinically relevant as they highlight the extent to which stroke survivors may adopt long-term compensatory strategies in the absence of appropriate orthotic intervention or when previous orthotic management has been unsuccessful. During the post-assessment recovery period, both participants reported prior experience with AFO provision but dissatisfaction with previous orthotic outcomes, including concerns relating to comfort, perceived functional benefit, footwear compatibility, and usability within daily life. This reflects a wider challenge within stroke orthotic rehabilitation whereby successful orthotic intervention is influenced not only by biomechanical correction, but also by patient acceptance, comfort, adherence, aesthetic considerations, and integration into everyday functional activities.

Although these broader orthotic and rehabilitation considerations were outside the direct scope of the current study, they provide important contextual insight into the complexity of the intended target population. The findings reinforce that gait deviation alone does not necessarily predict orthotic use or acceptance, and that clinical decision-making surrounding AFO provision remains highly individualised and multifactorial. Furthermore, the presence of compensatory movement strategies within participants not currently using

orthotic intervention highlights the potential value of combining biomechanical gait assessment with wider patient-reported and functional outcome measures in future work.

8.7 Summary

The findings presented in Chapter 8 confirm that it is feasible to deploy an integrated, multi-modal gait assessment protocol that simultaneously captures kinematics, neuromuscular activation, functional performance, and patient-reported outcomes in both neurologically intact and stroke-affected participants. Establishing this feasibility was central to the aims outlined in Chapter 3, and the successful implementation demonstrates that reliable, synchronised data can be obtained across multiple domains without undue burden on participants. This is a critical step, as clinical gait assessment too often isolates mechanical or neuromuscular variables, neglecting the interaction between motor control, functional capacity, and subjective experience. By embedding all four elements into a single protocol, the study provides a more holistic view of device action and participant response.

Critically, the results illustrate both the strengths and limitations of the tested devices. The fixed-ankle AFO promoted more physiological timing of gastrocnemius activity and stabilised shank-to-vertical progression, yet amplitude deficits persisted, reflecting the underlying neuromuscular impairments of stroke. The compression sock was well tolerated initially but comfort declined across repeated exposures, particularly in stroke participants. These findings highlight a recurring theme in orthotic and supportive technologies: interventions may achieve biomechanical normalisation but remain constrained by patient acceptance and long-term tolerability (Rosenberg et al., 2013; Partsch & Mortimer, 2015).

The juxtaposition of objective improvements with subjective decline underscores the importance of evaluating devices not only for efficacy but also for their real-world usability, an issue repeatedly identified in rehabilitation research as a barrier to sustained adoption.

Gait speed findings provide further context. Neurologically intact participants performed within normative ranges for young adults (Bohannon, 2011; Studenski et al., 2011), validating the accuracy of the measurement protocol. Stroke-affected participants, by contrast, walked at the lower threshold of community ambulation (Perry et al., 1995; Bowden et al., 2008).

This positioning is clinically significant: individuals at the margins of community ambulation may manage basic mobility but remain vulnerable to fatigue, falls, and environmental challenges. The absence of marked speed differences between the stroke groups with and without AFOs suggests that stabilisation and timing improvements do not automatically translate into faster walking. This finding aligns with mechanistic reviews showing that speed gains after stroke arise from coordinated changes across propulsion, balance, and energy efficiency rather than isolated device effects (Wonsetler & Bowden, 2017). The implication is that orthoses and compression supports should be integrated into broader rehabilitation strategies, not expected to drive functional improvements in isolation.

A further critical point concerns the constraints of the study design. Small group sizes, particularly in the stroke+AFO subgroup, limited statistical power and prevented detection of subtle differences across the gait cycle. This limitation, however, does not diminish the methodological advance of demonstrating feasibility. Rather, it highlights the next necessary step: larger-scale trials that can test the robustness and generalisability of the observed trends, incorporate subgroup stratification (by lesion type, chronicity, spasticity profile), and detect clinically meaningful changes in function. Importantly,

powering future studies for the minimum clinically important difference in gait speed (Tilson et al., 2010) would ensure clinical as well as statistical relevance.

From a translational perspective, the findings speak directly to the design and refinement of supportive devices. Improvements in material comfort, adjustability of compression, and modular ankle control could help mitigate the comfort decline observed, particularly in stroke survivors with altered sensory function. Since the underlying hardware and software have been shown to operate reliably, the challenge now lies in product design that prioritises usability, simplicity, and low cost, in line with the needs of overstretched clinical services. This focus is reinforced by the questionnaire study, which indicated that clinicians and patients prioritise devices that are straightforward to fit, comfortable for extended wear, and affordable for widespread use. Without such considerations, even technically effective devices risk limited translation into practice.

In conclusion, Chapter 8 provides critical feasibility evidence that integrated gait assessment protocols are workable and clinically relevant, while also revealing the limitations of current device iterations. The results highlight the dual imperative of refining device design for comfort and usability while expanding research to larger cohorts to establish efficacy with sufficient statistical power. Future progression should therefore combine engineering refinements with scaled clinical trials, ensuring that device solutions are not only biomechanically effective but also acceptable, affordable, and implementable in busy clinical contexts.

Beyond their immediate feasibility and clinical relevance, the results in Chapter 8 raise wider questions about how gait assessment and device-based interventions should be conceptualised within stroke rehabilitation and routine practice. They expose the trade-offs between biomechanical optimisation, neuromuscular adaptation, functional outcomes, and user acceptability, while

also underscoring the methodological challenges of evaluating small, heterogeneous clinical samples.

These complexities require interpretation in light of the thesis aims, existing literature, and the practical realities of clinical deployment. The subsequent discussion chapter (Chapter 9) addresses these issues directly, situating the findings within the broader rehabilitation landscape and critically examining their implications for both research and practice.

Chapter 9 - Discussion

9.1 Overview of Findings

This thesis explored three linked problems in post-stroke gait rehabilitation: constrained access to detailed biomechanical assessment, variability in clinical practice, and the feasibility of low-cost wearable tools to extend objective gait measurement beyond laboratories.

The programme comprised:

1. A UK-wide survey of current practice (Chapter 5).
2. Development and bench validation of a textile EMG + IMU prototype (Chapters 6–7).
3. Feasibility testing with non-affected and stroke-affected participants (Chapter 8).

The survey confirmed wide heterogeneity in assessment pathways and a predominant reliance on observational gait analysis, with limited access to 3D gait laboratories - despite the acknowledged value of objective metrics (Winstein et al., 2016; see Sections 5.5–5.6). Building directly on those findings, the prototype was engineered and validated against Vicon and activPAL to establish signal plausibility for core spatiotemporal and phasic muscle measures (Sections 7.5–7.6).

Feasibility work then demonstrated that stroke survivors can tolerate the garment and that usable data can be captured outside the lab when setup is

streamlined, while highlighting known pain points (electrode–skin contact stability, motion artefact, data-handling burden; Sections 8.5–8.6).

Together, these studies suggest a pragmatic path forward: wearable systems cannot replace high-fidelity laboratory analysis, but—with appropriate analytics and workflow fit—they can democratise access to *actionable* gait information in routine care, addressing structural inequities identified earlier (Sections 1.3, 2.3.10, 5.6.2–5.6.3).

9.2 Integration of Findings Across Studies

9.2.1 Linking Clinical Need and Technological Development

The clinician survey presented in Chapter 5 exposed a fundamental contradiction at the heart of stroke rehabilitation: gait is consistently prioritised as a key marker of functional independence, yet the tools available for its assessment remain dominated by subjective observation. Nearly nine out of ten clinicians reported assessing gait regularly, but two-thirds relied on unaided visual inspection, and fewer than 5% had any access to advanced technologies such as gait laboratories. These figures are striking when set against international best practice, which increasingly emphasises objective quantification of motor recovery (Kwakkel et al., 2019). The persistence of such a reliance on observational judgement is not simply a matter of tradition but reflects structural barriers in service provision. Time pressures, lack of trained personnel, and poor integration of biomechanics expertise into multidisciplinary teams were repeatedly cited in survey responses, suggesting that the problem is systemic rather than individual.

The implications of this gap are far-reaching. Current guideline frameworks, such as those issued by NICE (2013, 2019b), promote early and intensive

rehabilitation but remain largely agnostic about the measurement tools used to track progress. This omission tacitly legitimises the continued reliance on subjective observation. Yet the survey findings show that clinicians themselves are not content with this status quo: the majority expressed both dissatisfaction with existing methods and willingness to adopt portable, objective alternatives if these were affordable and integrated into routine workflows. This disconnect between guidelines and ground-level practice reveals an urgent need for translational innovation. Unlike many technology-focused studies, which begin with engineering possibilities and only later consider clinical application, the present thesis reverses this sequence: it begins by evidencing a clinical demand and then directs device development toward meeting it.

This grounding in real-world demand differentiates the thesis from much of the wearable sensor literature. Studies validating IMUs or textile EMG systems often demonstrate accuracy in laboratory contexts (Mancini et al., 2011; Kobsar et al., 2020), but without embedding their work in clinician-reported barriers to adoption. Consequently, many devices fail to progress beyond proof-of-concept because they do not address the pragmatic challenges of service delivery. By contrast, the design specifications developed here—rapid donning, minimal calibration, outputs tied to interpretable variables such as stance-time asymmetry and muscle onset timing—were explicitly informed by the survey results. This positions the thesis as not only a technical contribution but also an epistemological intervention, reorienting research from laboratory-driven to clinician-driven priorities.

Critically, this approach also challenges a common critique in rehabilitation technology: that devices represent “solutions in search of problems.” The evidence collected in Chapter 5 demonstrates that the problems are well defined by clinicians themselves: inadequate objectivity, lack of tools for orthotic tuning, and inequitable access to biomechanics expertise. The device development pathway in this thesis can therefore be seen as a direct response

to these gaps, rather than an imposition of engineering novelty onto clinical practice. In doing so, the work advances a broader argument: that rehabilitation technologies must be co-designed with, and justified by, the contexts in which they will ultimately be deployed.

9.2.2 Complementary Insights from Laboratory Validation and Real-World Feasibility

The laboratory validation reported in Chapter 7 confirmed that the prototype device was capable of delivering clinically meaningful metrics under controlled conditions. IMU-derived spatiotemporal parameters such as cadence and stride time achieved intraclass correlation coefficients above 0.80 when compared with motion capture, while EMG timing outputs remained stable across repeated trials provided that electrode placement and garment compression were consistent. These results place the device in line with the wider validation literature (Trojaniello et al., 2014; Kobsar et al., 2020), but more importantly, they invite a reconsideration of what constitutes “success” in wearable device validation.

Traditional studies prioritise amplitude fidelity and waveform morphology, often treating any deviation from laboratory gold standards as evidence of inadequacy. However, the findings here suggest a different framing: in rehabilitation practice, it is the *temporal structure* of muscle activation—onset, offset, and coordination with gait events—that carries clinical relevance. Amplitude variability, while scientifically noteworthy, does not necessarily undermine interpretability for orthotic decision-making.

This reframing is particularly salient when placed alongside the findings of Chapter 8, which assessed the device under real-world conditions. Here, expected challenges emerged: sweat, garment slippage, and ambient noise all compromised signal quality relative to the laboratory environment. Yet even

under these constraints, key features such as tibialis anterior activation during swing and gastrocnemius timing at toe-off were identifiable. The significance of this cannot be overstated. It demonstrates that while laboratory precision may be attenuated in clinical settings, the core outputs required for orthotic tuning and gait progression remain robust. In other words, the “accuracy ceiling” defined by laboratory testing is not eroded to the point of uselessness in practice; rather, the device maintains a “clinical floor of usability” that makes it viable where existing methods are weakest—routine care settings with no access to gait laboratories.

By juxtaposing laboratory and feasibility data, the thesis develops a layered argument about translation. The laboratory study demonstrates that the device meets standards of technical validity expected in biomechanics research. The feasibility study then demonstrates that these standards are not rendered irrelevant by ecological constraints but can still yield clinically actionable information. Together, these findings contest the binary framing often seen in the literature, which treats laboratory fidelity and clinical usability as mutually exclusive. Instead, the thesis argues that the value of wearable technologies lies precisely in their ability to balance both demands, achieving “good enough” precision to inform decision-making while remaining practical enough for everyday clinical use.

This integrative perspective also exposes limitations in the wider literature. Studies that focus exclusively on laboratory validation (e.g., Godfrey et al., 2015) risk producing devices that never leave research settings. Conversely, studies reporting only clinical feasibility (e.g., Lau et al., 2014) often lack the rigor needed to secure credibility in biomechanics. By deliberately combining both approaches, the thesis presents a coherent model for rehabilitation technology evaluation, one that can withstand scrutiny both from engineers and clinicians. The argument advanced here is therefore not simply that the

device “works,” but that it exemplifies a methodology by which wearable technologies should be developed, tested, and justified.

9.2.3 Feasibility Evidence from Chapter 8

The feasibility evidence presented in Chapter 8 adds a critical dimension to the overall evaluation of the system. Whereas earlier chapters (particularly Chapter 7) demonstrated the technical validity of the shank-to-vertical angle (SVA) approach in a controlled cohort, Chapter 8 established that it was possible to collect synchronised kinematic, EMG, gait speed, and patient-reported outcomes across both neurologically intact and stroke-affected participants, including those using ankle-foot orthoses (AFOs).

This is important because feasibility is often overlooked in technology-driven rehabilitation research: many devices function well in the laboratory but fail to integrate into real-world clinical workflows (Pizzolato et al., 2017). Demonstrating that complete datasets could be acquired without adverse events shows that the platform meets the first test of clinical translation — practical deployment in populations with motor impairments.

The results from Chapter 8 highlight the inherent trade-offs between methodological rigour and clinical realism. On the one hand, the EMG analyses confirmed clear between-group differences in gastrocnemius activation: non-stroke participants exhibited higher amplitude ($\sim 100\%$ RefMVC) with peak timing at $\sim 57\%$ gait cycle (GC), whereas stroke participants showed $\sim 40\%$ RefMVC with a delayed peak at $\sim 72\%$ GC. The stroke+AFO group demonstrated amplitude deficits similar to stroke alone, but timing was shifted closer to the non-stroke norm ($\sim 56\%$ GC). These results align with the wider literature demonstrating impaired plantarflexor recruitment after stroke (Mohan et al., 2021) and the timing-normalising effects of orthoses (Ohata et al., 2011).

However, despite clear descriptive patterns, the statistical analyses did not yield contiguous %GC ranges surviving FDR correction. This reflects the small sample sizes, especially in the AFO subgroup ($n = 2$), and highlights the challenge of balancing exploratory feasibility testing with the need for statistical robustness. As others have argued, early-phase device trials are vulnerable to Type II errors when sample sizes are insufficient to capture inter-individual variability (Dobkin & Dorsch, 2013). What matters in context is not the lack of significant clusters per se, but the demonstration that the device can elicit signal changes consistent with theory, which justifies progression to larger, confirmatory studies.

9.2.4 A Coherent Narrative of Innovation and Translation

When viewed across its phases, the thesis represents more than a collection of discrete studies: it forms a structured translational pipeline. Beginning with a needs assessment (Chapter 5), moving through device design and laboratory validation (Chapters 6–7), and culminating in feasibility testing under real-world constraints (Chapter 8), the research aligns closely with the staged approach advocated by the MRC framework for complex interventions (Craig et al., 2008) and the IDEAL-D guidelines for medical device development (McCulloch et al., 2013). This sequencing matters. Rehabilitation technology research is often fragmented, with devices tested opportunistically and without clear progression from proof of concept to clinical application. By contrast, the present thesis demonstrates that systematic staging is not only possible but also essential for generating evidence that is both scientifically credible and clinically persuasive.

This structure also enables a sharper argumentative stance about the purpose of wearable gait assessment. Too often, devices are judged against the standard of whether they can replicate full gait laboratory outputs. This sets an

unrealistic benchmark and implicitly devalues technologies that could transform practice in less glamorous but more impactful ways. The thesis argues that such comparisons miss the point. Wearable devices are not intended to *replace* gait laboratories but to *extend* their reach, making mechanistic gait assessment possible in contexts where it is otherwise absent. By reframing success in terms of accessibility, frequency of use, and clinical interpretability, the research shifts the evaluative criteria from precision as an end in itself to translational impact as the true measure of value.

Importantly, this reframing is not speculative but directly supported by the evidence gathered. Chapter 5 showed that clinicians overwhelmingly lack access to gait laboratories yet expressed willingness to adopt portable tools if they could deliver interpretable outputs. Chapter 7 demonstrated that the prototype could produce accurate spatiotemporal and temporal EMG measures under laboratory conditions. Chapter 8 confirmed that these outputs remained usable in clinical-like environments, despite ecological constraints.

Taken together, this body of evidence establishes not only that wearable gait analysis is technically feasible but also that it is responsive to explicit clinical demand. This trajectory transforms the thesis into a case study of translational logic: identifying need, developing technology, validating in the lab, and testing feasibility in practice.

This emphasis on clinically interpretable metrics also aligns with a broader and well-documented challenge within allied health practice: the inconsistent and often informal use of outcome measures. Recent evidence suggests that, although the majority of UK allied health professionals report using outcome measures, their application is highly variable, with many relying on informal interpretation rather than structured or validated frameworks (Young & Bowers, 2024). This issue is particularly pronounced within prosthetics and orthotics, where published evidence of routine, standardised outcome

measurement remains sparse. Barriers including time constraints, limited training, and lack of integration into clinical workflows further restrict meaningful use (Young & Bowers, 2024).

This gap is especially important in Prosthetics and Orthotics, where outcome measurement has historically been less embedded in routine practice compared to physiotherapy or occupational therapy. In this context, the present work can be understood not simply as a technological contribution, but as an attempt to operationalise outcome measurement within routine clinical practice. By reducing complex biomechanical data to a small number of clinically meaningful, interpretable indicators the system directly addresses known barriers to adoption. Rather than adding to the measurement burden, it seeks to embed outcome assessment within existing clinical workflows, thereby supporting the broader shift toward evidence-informed, data-driven rehabilitation practice.

Finally, this integrative argument highlights a broader philosophical point. Rehabilitation technology should not be judged solely by whether it advances biomechanical science; it must also be evaluated by whether it addresses inequities in care delivery. The thesis demonstrates that wearable devices can serve as tools of equity, bridging the gap between specialist infrastructure and routine practice.

By reorienting evaluation toward this criterion, the work contributes to a shift in how success is defined in the field: from the pursuit of laboratory-level fidelity to the pursuit of clinically meaningful accessibility.

9.3 Methodological Reflection

9.3.1 Staged Research Design and Philosophical Positioning

The methodological trajectory of this thesis, clinician survey, laboratory proof-of-concept, and feasibility testing, was designed to maximise translational value while retaining scientific rigour. This staged approach reflects not only the Medical Research Council (MRC, 2006; Craig et al., 2008) framework for developing and evaluating complex interventions but also embodies a philosophical stance that privileges pragmatism and applied realism. Unlike studies that begin with tightly controlled clinical trials and risk producing elegant but clinically irrelevant findings, this thesis deliberately built upwards from the realities of current practice. The decision to begin with a survey was not simply pragmatic but epistemological: it acknowledged that knowledge about gait assessment and rehabilitation does not reside solely in laboratory measures or statistical outputs, but in the daily practices and constraints of clinicians working at the front line of stroke care.

Such an approach reflects a pragmatist epistemology, where the value of knowledge is judged by its capacity to solve real-world problems rather than by its abstraction from context (Biesta, 2010). From this perspective, the integration of heterogeneous methods, qualitative insights from clinicians, quantitative biomechanical testing, and feasibility evaluation, was not a weakness but an essential strategy. Critics might argue that methodological pluralism risks diluting scientific precision. However, as Guba and Lincoln (1994) argue, ontological and epistemological commitments must match research purpose. Here, the purpose was translational: to generate evidence that would inform the design, validation, and adoption of wearable gait technologies in stroke rehabilitation. Methodological pluralism was therefore both philosophically defensible and practically necessary.

This staged, mixed-methods design also offered a safeguard against a common pitfall in technology-led rehabilitation research: technological determinism. Too often, devices are engineered in isolation from clinical need, with research questions shaped retrospectively around what the technology happens to measure (Godfrey et al., 2015). By embedding clinical insight at the outset and revisiting it throughout, the thesis created a feedback loop between need, theory, and evidence. This structure does not merely reflect the MRC framework but extends it, modelling how iterative engagement with stakeholders can strengthen not just the acceptability but also the conceptual coherence of emerging interventions.

9.3.2 Clinician Survey as Foundational Insight

The clinician survey (Chapter 5) was the first stage in this process, serving to identify service gaps and clinical priorities in UK stroke rehabilitation. With 66% of clinicians relying primarily on subjective observation and only 3% reporting access to gait laboratory facilities, the survey revealed a stark disconnect between the biomechanical complexity of hemiparetic gait (see Section 2.3) and the tools available to capture it. Time (72%) and staffing (54%) were reported as greater barriers than cost (41%), suggesting that clinicians are not resistant to technology per se but constrained by workload and service design.

These statistics were not only descriptive but formative: they justified the later methodological focus on portable, low-cost, and rapid-to-deploy technologies. In isolation, the survey could be critiqued for limited inferential power and for being conducted during a period of pandemic disruption, when service patterns were atypical. Yet its credibility rests on triangulation with existing literature. Tyson and Connell (2009) similarly reported that gait assessment in stroke rehabilitation remains dominated by subjective visual observation, while Pollock et al. (2014) highlighted the persistent underutilisation of instrumented

measures in UK practice. By situating the survey findings within this broader evidence base, the thesis mitigates the risk of local bias and strengthens the generalisability of the insights.

More importantly, the survey functioned as a methodological counterweight to the laboratory studies. Where the latter risked privileging fidelity and control, the survey emphasised feasibility and context. This dual perspective directly influenced the decision to emphasise metrics such as the shank-to-vertical angle (SVA) and EMG onset timing, which are interpretable by clinicians and tunable within orthotic practice (see Section 2.4), rather than complex kinematic parameters requiring laboratory infrastructure. In doing so, the survey was not a methodological “soft start” but a decisive shaping force. Without it, the thesis risked drifting into the technological determinism critiqued above. Instead, the survey ensured that methodological choices were anchored in clinical demand, exemplifying the co-design ethos increasingly promoted in rehabilitation engineering (Peters et al., 2020).

9.3.3 Proof-of-Concept Testing under Pandemic Constraints

The proof-of-concept study (Chapter 7) was conducted at a moment of acute methodological constraint. COVID-19 restrictions prevented recruitment of stroke survivors, forcing reliance on a homogenous sample of healthy volunteers drawn from the university community. On first inspection, this appears a limitation: results from healthy participants cannot capture the asymmetries, spasticity, and fatigue-related variability characteristic of hemiparetic gait. Yet when judged by the standards of proof-of-concept research, the design was entirely appropriate. As Kistin and Silverstein (2015) and Leon, Davis, and Kraemer (2011) note, the purpose of proof-of-concept studies is not to generalise to clinical populations but to establish whether an intervention—or in this case, a device—is technically feasible and safe enough to justify progression.

By adopting this approach, the study could address foundational questions: Could textile-based EMG electrodes provide stable, interpretable signals? Could the activPAL reliably capture shank-to-vertical angle (SVA) during gait? Were donning and doffing procedures practical for repeated use? These questions, while modest, were essential. Attempting to answer them in a stroke population during the pandemic would not only have been ethically questionable but methodologically muddled, as variability in hemiparetic gait would have obscured device-level signal fidelity. In this sense, the homogenous cohort acted as a methodological “clean room,” isolating technical performance from clinical complexity.

Of course, this design carried trade-offs. Findings could not speak to the device’s sensitivity to pathological gait patterns, nor to its usability in populations with fatigue, spasticity, or cognitive impairment. Yet this limitation was openly acknowledged and explicitly corrected in the subsequent feasibility study (Chapter 8). Here lies a critical strength: the staged design did not deny its compromises but absorbed them into a broader research trajectory. In doing so, it echoed Watermeyer et al.’s (2021) observation that pandemic-era research required adaptive methodologies that prioritised both continuity and caution. Rather than being a methodological weakness, the proof-of-concept design exemplified adaptive rigour: it preserved the research pipeline under exceptional circumstances while ensuring that subsequent phases could proceed on a validated technical foundation.

A further limitation relates to the extent of co-design within the development process. While the survey in Chapter 5 ensured that clinician perspectives directly shaped device priorities, formal involvement of stroke survivors during the early design phases was not undertaken. This reflects a pragmatic constraint of the staged research design, in which proof-of-concept validation necessarily preceded user-centred refinement.

However, it also represents a limitation. Service user involvement is increasingly recognised as essential in rehabilitation engineering, not only to enhance usability and acceptability but to ensure that technological solutions align with lived experience and meaningful functional outcomes. The absence of structured patient and public involvement at the design stage may therefore have limited the extent to which user priorities, such as comfort over extended wear, ease of donning, and perceived benefit, were fully embedded from the outset.

Importantly, this limitation is not conceptual but temporal. With technical feasibility now established, future phases of this work will prioritise co-design approaches involving stroke survivors as active partners in device refinement. This will be particularly critical as the system evolves toward greater clinical usability, where factors such as wearability, interface design, and perceived value become as important as signal fidelity. Embedding service user perspectives at this stage is expected to strengthen both translational potential and real-world impact.

9.3.4 Feasibility Testing and the Balance of Rigour and Realism

The feasibility study (Chapter 8) represented the methodological culmination of the thesis, testing the wearable system in a stroke population. Feasibility studies often attract criticism for small sample sizes, limited statistical power, and lack of definitive outcomes (Arain et al., 2010). Yet such critiques miss their purpose. As Eldridge et al. (2016) emphasise, feasibility designs are intended to answer practical questions about recruitment, acceptability, safety, and signal reliability, not to produce generalisable effect sizes. In this thesis, those questions were decisive. Could stroke survivors tolerate a compression sock with integrated electrodes? Would hemiparetic gait patterns compromise EMG signal fidelity? Could SVA be meaningfully extracted from activPAL

outputs in a clinical cohort? The study showed that, despite heterogeneity of impairment, these aims were achievable.

What gives this feasibility study particular weight is its explicit linkage to both earlier stages and wider literature. The survey (Chapter 5) had revealed clinicians' appetite for portable, objective gait measures, but scepticism about laboratory-heavy protocols. The proof-of-concept study (Chapter 7) had demonstrated technical viability under controlled conditions. The feasibility study therefore functioned as the bridge: it tested whether those technical gains could be realised in the population that mattered most. In this respect, it stands as a methodological exemplar of the "scaffolding" approach advocated by Eldridge et al. (2016), where early-phase work is not dismissed as inconclusive but valued as essential groundwork for later trials.

Critically, the feasibility results can be read against systematic literature gaps identified in Chapter 2. For instance, propulsion-targeted gait training studies often excluded orthotic users (Burpee & Lewek, 2015), raising questions about ecological validity. The feasibility study here directly countered such exclusions by incorporating participants who reflected real-world clinical practice, including those using orthoses. In doing so, it provided empirical evidence that the wearable device could capture gait parameters even in the presence of orthotic variability. This methodological inclusivity strengthens the translational potential of the work, offering a contrast to trials whose methodological "purity" undermines clinical relevance.

Finally, the feasibility design highlights the tension between rigour and realism in rehabilitation research. Large RCTs offer rigour but often exclude the very patients most in need of innovation. Feasibility studies, by contrast, embrace realism but sacrifice statistical certainty. The methodological contribution of this thesis lies in showing how these poles can be balanced across a staged trajectory. By acknowledging limitations at each stage, yet ensuring they were

counterbalanced in subsequent designs, the research provides not only empirical findings but also a methodological template for translational rehabilitation science.

9.4 Gait Analysis Accessibility and Equity

A central tension highlighted by this thesis is the contrast between the clinical value of detailed gait analysis and the limited opportunity most stroke survivors have to benefit from it in practice. The survey of UK clinicians (Chapter 5) made this explicit: 66% of respondents reported relying primarily on observational gait analysis, while only 3% described access to gait laboratory facilities (Sections 5.5–5.6). These figures are not merely descriptive but expose a systemic inequity. While research and specialist centres have demonstrated the clinical significance of parameters such as propulsion, trailing limb angle, muscle activation timing, and shank-to-vertical angle (SVA) (Clark et al., 2010; Kerr et al., 2020), such mechanistic insights remain largely invisible in routine care. This creates a two-tier system of rehabilitation: one grounded in evidence-based biomechanical detail, the other dependent on subjective clinical impression. The inequity is not simply practical but epistemological, shaping what knowledge is available to guide orthotic tuning, therapy progression, and prognostic decision-making.

Geographic distribution of services is one of the clearest determinants of this inequity. Gait laboratories are clustered within major teaching hospitals, leaving vast catchments underserved (Section 5.5.5). Similar centralisation has been reported internationally, with Baker (2013) describing it as a structural feature of high-income health systems, while in low- and middle-income countries gait laboratories remain virtually absent. The impact of this distribution is compounded by the nature of the stroke population. Travel to tertiary centres imposes a disproportionate burden on those already

contending with fatigue, mobility restrictions, and reliance on caregivers. Evidence from the NIHR (2018) and Kelly et al. (2016) demonstrates a clear distance–decay effect in service uptake, with attendance falling sharply as travel requirements increase. Survey respondents in this thesis confirmed this problem, describing cases where patients missed or declined gait lab appointments because the logistical effort outweighed the perceived benefit (Section 5.6.3). When combined with the social gradient of stroke incidence and disability (Marmot, 2020), these barriers ensure that the very populations most likely to require detailed gait assessment are least likely to access it. The inequity is therefore doubly compounded: stroke is more prevalent in disadvantaged groups, and services to address its functional consequences are systematically less available to them.

Yet geographic inequity is only one part of the problem. Even when access is theoretically available, the procedures of gait laboratories may not be well suited to stroke populations. Standard 3D gait analysis (3DGA) protocols involve marker placement, calibration, synchronisation with force plates and EMG, and repeated walking trials to ensure repeatability. These requirements demand concentration, balance, and endurance. For survivors with hemiparesis, spasticity, cognitive impairment, or fear of falling, such procedures can be intolerable. Del Din et al. (2016) and others have critiqued laboratory workflows for prioritising methodological precision at the expense of patient suitability.

The survey findings in this thesis mirror these concerns: clinicians reported that the very patients most in need of high-fidelity biomechanical analysis—those with severe gait impairments—were often those least able to complete laboratory sessions (Section 5.6.3). The result is a form of selection bias, where those with milder impairments are systematically over-represented in both clinical assessments and the published literature. In practice, this means that orthotic prescriptions may be tuned based on visual estimates of SVA

rather than measured parameters, and therapy progression may be decided without quantitative spatiotemporal data. The implications extend beyond access to the very structure of evidence: datasets that inform intervention studies risk being skewed towards higher-functioning patients, leaving gaps in guidance for those with complex needs.

A third barrier lies in workforce capacity and expertise. Interpreting multidimensional biomechanical and neuromuscular data requires skills that are concentrated in a limited number of specialist centres. The survey findings (Sections 5.6.5–5.6.7) demonstrated that many clinicians lack confidence in using or interpreting such data, a concern reinforced by national workforce reports that list physiotherapists, orthotists, and occupational therapists as shortage occupations (UK GOV, 2023). This shortage means that even when data are available, they may not be translated into meaningful clinical decisions.

The broader literature on digital health echoes this concern, with Hug et al. (2021) warning that excessive data complexity risks overburdening clinicians rather than aiding them. Raw EMG traces or kinematic plots may offer rich scientific information but can overwhelm clinicians working in time-pressured services. The solution, as argued in Chapters 7 and 8, lies not in reproducing the analytic complexity of gait laboratories but in distilling data into clinically parsimonious, validated indicators such as stance-time asymmetry, tibialis anterior onset timing, and SVA alignment. Such measures sacrifice some granularity in favour of interpretability but offer a viable route for redistributing analytical capacity across the broader rehabilitation workforce. The balance between fidelity and usability is therefore central to the equitable adoption of gait technologies.

Any discussion of accessibility must also address the digital divide. Populations at greatest risk of stroke, older adults, socioeconomically deprived

groups, and certain minoritised ethnic communities, are also those most likely to face barriers to digital engagement (George et al., 2020). Without careful planning, the introduction of wearable technologies risks reproducing the “inverse care law” described by Greenhalgh et al. (2017), whereby innovations first benefit those who need them least. The survey data show a more complex reality. While 82% of clinicians expressed willingness to adopt portable, low-cost technologies (Section 5.6.7), concerns were raised about patient usability, clinician training, and integration into existing care pathways.

These findings are consistent with broader literature on digital rehabilitation, which shows that innovations can exacerbate inequities unless accompanied by accessible interfaces, multilingual support, and non-digital fallback options. The prototype device developed in this thesis was designed with these concerns in mind: its emphasis on comfort, simplicity, and integration into established clinical routines reflects a commitment to equity in both design and deployment. Yet technology alone cannot resolve disparities. Unless wearable devices are embedded within care pathways in ways that explicitly address digital exclusion, there remains a risk that new tools will reinforce rather than mitigate inequities.

Taken together, these barriers—geographic distribution, procedural suitability, workforce capacity, and digital exclusion—demonstrate that inequitable access to gait analysis is not an incidental flaw but a structural feature of current rehabilitation provision. The evidence from both the literature and the empirical work in this thesis indicates that the populations most likely to need mechanistic gait analysis are systematically excluded by geography, by protocol, by workforce shortages, and by digital barriers. Addressing this requires more than incremental improvement to existing laboratory systems. Instead, it necessitates a shift in paradigm: from centralised, resource-intensive, laboratory-based assessments towards portable, interpretable, and clinically integrated systems that can extend mechanistic insight to the point of

care. The wearable solutions trialled in this thesis (Chapters 7–8) exemplify one such approach, offering evidence that core gait parameters can be captured outside the laboratory without imposing excessive burden on patients or clinicians.

This is not an argument to replace gait laboratories, which remain essential for complex surgical planning and detailed kinetic modelling (Baker, 2013), but to complement them within a tiered system of assessment. In such a system, high-fidelity laboratory analyses are reserved for complex cases, while portable technologies extend access to objective measurement for the wider stroke population, thus narrowing the inequities that currently characterise gait rehabilitation.

9.5 Critical Analysis of Wearable EMG/IMU Development

The technical contribution of the present study is the development of a fabric-integrated sensing system combining electromyography (EMG) with a shank-mounted inertial measurement unit (IMU). Its novelty lies in the deliberate trade-off between laboratory fidelity and ecological feasibility, shaped by survey findings that clinicians prioritise rapid set-up, interpretability, and comfort over maximal precision (Ch. 5, Sections 5.6.7–5.6.8). The system is positioned not as a competitor to 3D gait laboratories or robotic trainers, but as a pragmatic response to an identified service gap: the lack of objective, repeatable, and affordable gait data in everyday stroke rehabilitation practice.

9.5.1 Novelty, Validation, and Usability

Chapter 8 reinforced the novelty of the integrated EMG/IMU set-up by showing that reliable neuromuscular signals could be obtained even from stroke

participants with spasticity and altered motor control — populations in which EMG signals are notoriously noisy and often rejected from analysis (Burden, 1999). The ability to record, normalise, and interpret these signals demonstrates technical validity and robustness. Importantly, the inclusion of VAS scores confirmed that usability extends beyond technical function: compression devices were feasible and safe, but progressive declines in comfort raised usability concerns. This mirrors broader trends in orthotic adoption, where technical promise is undermined by poor patient tolerance (Ramstrand & Lygnegård, 2012).

The rationale for embedding EMG into a textile platform is grounded in a long-standing tension between fidelity and feasibility. Conventional Ag/AgCl electrodes remain the “gold standard” due to their low impedance and high signal quality (Merletti & Farina, 2016), but they impose preparation burdens, risk drying artefacts, and are disliked by many patients for their residue and need for repeated skin preparation (Bonato, 2010). The clinician survey echoed these complaints, with respondents citing time and tolerability as major barriers to wider EMG use (Ch. 5). Textile electrodes, stitched into compression garments, directly address these barriers but bring their own challenges: variability in signal stability, dependence on electrode–skin contact pressure, and susceptibility to motion artefact (Catrysse et al., 2004). In the system reported here, these trade-offs were addressed through compression zoning and optimised electrode footprints (Ch. 6), representing a synthesis rather than a rejection of conventional methods.

The decision to integrate IMU data is not merely additive but transformative. By anchoring EMG activity to gait cycle events, the system redefines the value proposition of wearable EMG: shifting from amplitude-driven fidelity, which is highly variable across sessions and garments, to timing-driven features, which are both more stable and more clinically relevant. This reflects evidence from Clark et al. (2010) and Awad et al. (2016), who demonstrated that impaired

temporal coordination, such as delayed tibialis anterior onset or abnormal gastrocnemius co-activation, is more predictive of propulsion deficits and walking inefficiency than amplitude anomalies. The validation work confirmed this shift: temporal variables such as step time and activation onset were robust, while amplitude features fluctuated (Ch. 7). This supports a reframing: textile EMG should not be judged against the unattainable standard of laboratory amplitude accuracy but rather evaluated on its ability to reliably capture clinically meaningful timing anomalies.

Comparative evidence strengthens this argument. Mancini et al. (2011) and Kobsar et al. (2020) demonstrated intraclass correlations $>.80$ for cadence and stride time when IMUs were compared with motion capture, while textile EMG remains controversial for amplitude fidelity (Merletti & Farina, 2016). By demonstrating acceptable agreement for timing-based EMG features and spatiotemporal IMU outputs, the results show that the perceived weaknesses of textile EMG are manageable when contextualised by multimodal sensing. This challenges the common critique that textile EMG is “too noisy” for clinical use, by showing that “noisy” amplitude measures are not the features clinicians most need.

The usability findings push this comparative argument further. Patients tolerated the compression garment well, preferring it to gel electrodes and reporting comfort when compression was adjusted appropriately (Ch. 8). Issues of slippage, perspiration, and donning variability remain, but these are surmountable design refinements—silicone grippers, vented fabrics, and calibration prompts—rather than insurmountable flaws (Tabor et al., 2018). Critically, clinicians emphasised the importance of speed and simplicity in set-up (Ch. 5), reinforcing Greenhalgh et al.’s (2017) claim that adoption barriers are more often organisational than technical. By directly addressing these priorities, the device situates itself as a credible solution to a documented implementation gap.

Taken together, the novelty of the work reported here is not reducible to its components—textile EMG and IMUs are both well described in the literature—but lies in the reframing of their limitations. Where others see “unreliable amplitude,” this research demonstrates “robust timing.” Where others see “reduced fidelity,” this research shows “increased feasibility.” By aligning technical compromises with clinical priorities, the system challenges the orthodoxy that only high-fidelity laboratory tools are valuable, advancing an alternative model where the benchmark is not the laboratory, but routine clinical need.

9.5.2 Translational Positioning and Systemic Challenges

The true test of innovation lies in translation. Here, the system must be situated within the broader rehabilitation technology landscape and critically compared to existing alternatives. Robotic gait trainers, for example, deliver pooled relative risks for improved walking independence of 1.45 (95% CI 1.22–1.72; Mehrholz et al., 2017). Yet they remain prohibitively expensive, centralised, and accessible to only a minority of stroke survivors. Commercial IMUs such as activPAL offer robust step counts and spatiotemporal data (Mancini et al., 2011) but cannot provide neuromuscular insights. Laboratory-based gait analysis, meanwhile, provides unmatched fidelity but is geographically restricted and burdensome to patients (Chapter 5). In this comparative context, the present system deliberately situates itself in the middle ground: richer than step monitors, cheaper and more accessible than laboratories, and more targeted than robotics. Critics may dismiss this middle ground as “neither precise nor simple enough,” but the counterargument advanced here is that the middle ground is precisely where the unmet clinical demand lies.

The survey findings sharpen this point. Only 3% of clinicians in the UK reported access to gait labs, while 72% cited time as the major barrier to advanced gait

assessment (Chapter 5). Insisting on laboratory-grade accuracy as the standard of care is therefore not only impractical but inequitable: it restricts biomechanical insight to a small, urban minority and excludes those most in need. Marmot's (2020) work on health inequities underscores this point: tools must be judged not only on efficacy but also on distribution. By expanding reach at the expense of some precision, wearable EMG/IMU systems align more closely with equity-driven models of care than gold-standard but inaccessible alternatives.

This translational argument, however, cannot ignore systemic challenges. The prototype's reliance on Arduino-class electronics is acceptable in proof-of-concept stages but incompatible with regulatory and clinical deployment. To progress, medical-grade hardware, GDPR-compliant data storage, and CE/UKCA certification are mandatory (Holden et al., 2020). Critics may argue that such transitions will erode affordability. Yet, as the Medical Research Council (2006) emphasises, early-phase feasibility work is designed precisely to surface these tensions. Rather than undermining the project, these constraints mark the path from concept to implementation, highlighting where technical solutions must be matched by regulatory and economic ones.

The workforce capacity issue further complicates translation. Survey data revealed shortages of physiotherapists, occupational therapists, and orthotists (Chapter 5), raising the question of who will deploy and interpret wearable outputs. Patel et al. (2012) argued that digital health tools often fail not due to technical inadequacy but due to excessive complexity in clinical interpretation. The present research echoes that critique and offers a counter-model: "clinically compressed analytics" that distil raw signals into a small number of validated, interpretable metrics (e.g., stance-time asymmetry, activation onset delay). By aligning outputs with clinical decision points rather than raw data streams, the system mitigates the risk of overburdening staff and ensures translational viability.

Finally, the comparative position of the system must be clarified: it is not designed to replace gait laboratories, but to complement them. Baker (2013) and Del Din et al. (2016) emphasise that laboratories remain indispensable for complex surgical planning and atypical gait presentations. Yet, as documented in Chapter 5, the alternative for most stroke survivors is not laboratory access but reliance on subjective observation. Against that benchmark, the value of wearable EMG/IMU systems becomes clear: they democratise access to objective data, even if they cannot replicate the full fidelity of laboratory measures. This reframes the translational challenge not as a competition with laboratories, but as an ethical obligation to provide “good enough” tools where “perfect” tools are unattainable.

From a translational standpoint, the feasibility findings show that the system sits in an intermediate position between laboratory prototypes and clinically deployable tools. The hardware and software proved stable, yet the clinical impact of the devices was constrained by design factors (rigid ankle setting, compressive discomfort). This echoes prior warnings that technological innovation alone does not guarantee adoption; success depends on embedding solutions within patient-centred and workflow-compatible frameworks (Chen et al., 2020). The Chapter 8 data therefore strengthen the translational case, but also clearly map the barriers that must be overcome before real-world implementation is viable.

In sum, the translational critique positions the system as both scientifically defensible and ethically compelling. Its comparative advantage lies not in outperforming laboratories but in offering scalable, interpretable, and equitable access to objective gait data. By situating itself between high-fidelity but inaccessible tools and low fidelity but widely available ones, the system advances a new category of rehabilitation technology: one that prioritises pragmatic feasibility without abandoning mechanistic insight.

9.6 Comparisons with Existing Rehabilitation Technologies

The studies undertaken in this research collectively argue that a dual-modality, fabric-integrated EMG/IMU can change *how* objective gait information is generated and used in routine stroke rehabilitation. Rather than reheating the literature, the task here is comparative and critical: to evaluate what the present system adds—and where it still falls short—relative to the dominant options in practice (observation), the canonical gold standard (3D gait laboratories), high-intensity training systems (treadmills/robotics), and commercially mature inertial platforms. The discussion leans on findings reported in Chapters 5 (questionnaire, n=UK clinicians), 7 (proof-of-concept validation), and 8 (feasibility in stroke), and sets those results against the capabilities and constraints of existing technologies (Baker, 2013; Kirtley, 2006; Toro et al., 2003; Mancini et al., 2011; Mehrholz et al., 2017; Polese et al., 2013).

9.6.1 Observation, Laboratories, and Robotic Systems: Strengths and Systemic Failure Points Exposed by the Present Findings

Clinically, observation dominates because it is fast, universal, and practically cost-free. The survey in Chapter 5 showed 89% of respondents routinely assess gait, but 66% reported relying chiefly on visual inspection; only 3% had access to gait laboratory facilities, and constraints were consistently reported for time (72%), staffing (54%), and cost (41%). These are not marginal frictions: they define the operational envelope within which most decisions are made. The implication for technology is stark. Any solution that demands long set-up times, multiple operators, or specialist rooms will predictably miss most patients most of the time—irrespective of its scientific elegance. The present wearable was designed precisely against that constraint set: compression-

integrated electrodes to reduce skin prep; a single shank-mounted IMU to anchor temporal events; and summarised outputs that clinicians said they could act on (Sections 5.6.7–5.6.8). In other words, the design logic took the real clinical denominator—time, not technology—as the primary requirement. Yet an argument from practicality must not slide into an argument from complacency.

Visual assessment fails systematically when precision is needed most: subtle temporal lags, context-dependent asymmetries, and short-lived deviations at transition phases (e.g., late stance to pre-swing). Inter-rater agreement on complex patterns is modest (Toro et al., 2003), and the cognitive load of extracting multi-parameter judgements in real time is high (Kirtley, 2006). By embedding tibialis anterior (TA) and gastrocnemius timing relative to IMU-derived gait events, the present system moves the decision point from “what did I think I saw?” to “what changed and by how much?”—particularly for orthotic tuning and progression where the survey identified the greatest need for objectivity (Chapter 5.6.3). Importantly, validation (Chapter 7) confirmed that temporal indices were robust even when amplitude varied with compression or sweating—precisely the scenario in which observation is most brittle.

The comparison to 3D gait laboratories demands candour. Laboratories deliver detailed kinematics and kinetics, joint moments and powers, and synchronised EMG with face-valid precision (Baker, 2013). They remain indispensable for complex presentations, surgical planning, and atypical neuromechanics. The present system does not, and should not claim to, replicate those capabilities. What the findings suggest, however, is that the *modal* patient encounter does not require the full laboratory stack to extract actionable information. Chapters 7 and 8 showed that a short, bedside-deliverable protocol could index timing shifts in TA and gastrocnemius relative to stance/swing events and provide shank-to-vertical angle (SVA) estimates that clinicians can use to tune ankle—

foot orthoses (AFOs) without a gait lab visit. The survey's geography and capacity data (only 3% with lab access; Chapter 5.5) turn this from a convenience argument into one about equity: if laboratory access is the gatekeeper for objective data, then patients' postcodes and fatigue thresholds—not their clinical needs—determine who receives mechanistic assessment. The wearable contests that gatekeeping by lowering the bar for when and where useful signals can be captured.

Robotic and treadmill systems sit in an adjacent category. Meta-analyses report small-to-moderate improvements in speed and independence (e.g., absolute gains ~0.05–0.14 m/s; pooled RR $\approx$ 1.45 for independent walking in selected subgroups), but these platforms are centralised, staff-dependent, and designed to *train*, not to assess (Mehrholtz et al., 2017; Polese et al., 2013). The feasibility study in Chapter 8 made that divergence explicit: clinicians did not request more hardware to deliver repetitions; they requested objective markers that would make daily decisions safer and faster—whether to adjust SVA, progress velocity targets, or flag atypical co-contraction. In that sense, the wearable complements, rather than competes with, capital-intensive trainers: it can accompany patients to the ward, the community clinic, or the home, sampling brief episodes repeatedly to guide the dose and focus of conventional or robotic training between rarer lab visits.

There is a legitimate counterargument: if laboratory measures are the reference, does a lower-fidelity device risk normalising error? The validation choices in Chapter 7 confront that risk directly. The device was not evaluated against a single correlation coefficient; it was challenged on limits of agreement for temporal events and SVA estimates, with explicit reporting of cases where textile EMG amplitude drifted. The *design* response—fusing EMG with IMU to privilege timing over amplitude—addresses error structure rather than ignoring it. This is precisely the sort of methodological humility the field needs: acknowledging that the clinically useful subset of signals is not the

same as the statistically rich subset of signals, and engineering around that truth.

Finally, the thesis does not exempt the wearable from systems realities. Arduino-class prototypes are ideal for hypothesis-testing but must evolve to medical-grade electronics, secure data pipelines, and integration with electronic health records to scale (Holden et al., 2020; Patel et al., 2012). Nor does it claim that every clinical use-case is served by a two-channel EMG paired with a shank IMU; children with CP, complex multi-segment deformity, or surgical planning will continue to require laboratory-level analyses. The comparative claim is narrower and sharper: for the majority of routine decisions in adult post-stroke care—where 66% of clinicians currently depend on observation and 72% cite time as the binding constraint—the present system provides *different information* (activation timing anchored to gait events) at a *different cost profile* (minutes, not hours), in *different places* (bedside, clinic, home). That reframes the default pathway without displacing the pinnacle of precision.

9.6.2 Commercial IMUs and the Case for EMG–IMU Fusion: From Description to Explanation

Commercial IMUs (e.g., activPAL, Opal, Xsens) excel at what they were designed to do: detect steps, quantify cadence and stride time with high test–retest reliability, and classify posture and activity with validated algorithms (Mancini et al., 2011; Kobsar et al., 2020). In Chapter 7, the proof-of-concept comparisons reflected this maturity: cadence and stride timing agreed well with reference systems (intraclass correlations commonly $> .80$; typical errors often $< 5\%$), confirming that inertial timing is a solved engineering problem in controlled conditions. If the clinical question is *how much* someone walks or *how often* they transition, commercial IMUs are already fit-for-purpose and deployment-ready.

The present studies, however, foreground a different clinical question—*how* movement is produced—and here inertial data alone are mute. AFO tuning depends on understanding push-off and shank progression; propulsion-targeted therapy depends on the timing of plantarflexor activity relative to trailing limb angle; fall-risk management often hinges on late-swing dorsiflexor activation. Commercial IMUs can infer that an event occurred; they cannot say whether tibialis anterior was late, whether gastrocnemius activity was prolonged into early swing, or whether co-contraction patterns changed with speed. Chapters 7 and 8 demonstrate that adding even a *minimal* EMG channel set can answer these questions.

The dual-modality system mapped TA onset and offset to gait events and provided interpretable gastrocnemius timing relative to terminal stance. Where the textile electrodes introduced amplitude variability, the IMU anchored timing; where the IMU's event detection faced ambiguity (e.g., shuffling, brief hesitations), EMG phasing disambiguated. This is not redundancy; it is complementarity.

A legitimate critique is that textile EMG is fragile: contact pressure must be “in the pocket,” moisture changes impedance, and donning errors propagate to signal quality (Merletti & Farina, 2016). The thesis does not elude these issues; it designs into them. The compression zoning and electrode footprint (Chapter 6) targeted stable skin–electrode pressure; donning prompts and brief calibration routines reduced placement error; the analysis pipeline privileged *relative* timing features over absolute microvolts. Validation results were consistent with the broader literature: amplitude features varied more than timing features, but the latter were stable enough to be clinically interpretable (Chapter 7.6). In effect, the thesis argues that when the *clinical* decision is about timing (late TA, prolonged gastrocnemius), insisting on amplitude fidelity

as the primary criterion over-specifies the problem and mis-aligns the device with its use-case.

There is also a systems-level comparison to weigh. Commercial IMUs benefit from robust hardware, robust firmware, warranties, and support teams. The prototype here does not - yet. But the countervailing advantage is conceptual: EMG–IMU fusion converts inertial *description* into a mechanistic *explanation*. This mattered to respondents in Chapter 5: 82% expressed willingness to adopt portable, low-cost technologies *if* they delivered interpretable, clinically meaningful outputs, with orthotic assessment cited as a priority gap. A step count cannot tell an orthotist how to tune SVA; a TA onset index aligned to late swing can. A cadence value cannot reveal whether propulsion improved because trailing limb angle increased or because plantarflexor timing shifted; paired IMU–EMG can show the coordination change even when spatiotemporal metrics remain flat. This is the specific, function-linked informational value that commercial IMUs lack, and the present system provides.

Two further comparative points strengthen the argument. First, ecological validity: Chapters 7–8 showed the system working in short, low-fatigue bouts typical of bedside or community reviews, not only in laboratory corridors. That aligns with the *deployment* context clinicians described and where commercial IMUs often revert to long, unsupervised wear for activity profiling rather than targeted, coached assessments. Second, orthotic integration: because the system indexes shank orientation (SVA proxy) and muscle timing, it sits naturally within AFO review workflows, where changes can be made and re-measured immediately. Commercial IMUs can report that walking speed increased; the EMG–IMU can show *why*, which is the currency clinicians need for tuning.

The counterarguments deserve space. Textile durability remains a constraint (wash cycles, oxidation of conductive fibres), as do regulatory and data governance issues that scale beyond prototypes (Holden et al., 2020; Patel et al., 2012).

The thesis addresses these as next-stage engineering and translational problems, not as reasons to revert to inertial-only monitoring. More importantly, it resists the false dichotomy that an EMG–IMU must replace commercial systems. The comparative claim is additive: retain commercial IMUs where activity surveillance is the goal; deploy EMG–IMU when the clinical decision hinges on *coordination* rather than *quantity*. The studies here provide the empirical basis for that division of labour.

In sum, the critical comparison is not about higher numbers on a spec sheet; it is about informational fit to clinical purpose. Commercial IMUs are excellent at counting and timing; 3D labs are unparalleled at quantifying forces and joint powers; robots and treadmills are built to deliver dose.

The present work shows that a modest, low-burden EMG–IMU can occupy the under-served space where clinicians need fast, local, mechanistic insight to justify an orthotic adjustment, to explain a speed plateau, or to decide that a propulsion-targeted block is warranted—all in contexts where the alternative is still, too often, “look harder.” That is the comparative edge argued by the data in Chapters 5, 7, and 8 and the niche this system is designed to fill.

9.7 Cost–Benefit and Health Service Implications

The case for decentralised, fabric-integrated wearable sensing systems is not only technical but also economic and systemic. Stroke rehabilitation represents one of the heaviest recurrent costs in UK health and social care,

with estimates exceeding £26 billion annually, around half of which falls directly on health and community services (Stroke Association, 2023; Section 1.3). Yet despite this burden, the evidence presented in Chapter 5 highlights that most UK stroke services still rely on subjective gait observation, with only 3% of respondents reporting access to gait laboratories, and 72% citing time as a limiting factor in delivering detailed assessments.

These statistics underscore a paradox: the conditions that most demand precision and efficiency are those least able to access them. This mismatch creates a space for wearable systems that can deliver objective, mechanistic insight at the point of care, without incurring the prohibitive capital, staffing, and logistical costs of laboratory-based technologies.

The economic argument begins with burden concentration. Long-term disability, secondary musculoskeletal complications, falls, and recurrent admissions account for much of the lifetime cost of stroke care (Langhorne et al., 2011). The studies in Chapters 7 and 8 showed that even minimal indices—such as tibialis anterior onset timing or shank-to-vertical angle estimation—can provide interpretable, clinically actionable markers for orthotic tuning and therapy progression. If maladaptive gait patterns can be detected and corrected earlier through such portable assessments, the plausible downstream effects include reduced orthotic failure, fewer preventable falls, and improved endurance, each of which carries direct and indirect system savings. These potential benefits, however, are contingent not on technological promise alone but on the integration of wearable data into existing review and decision-making pathways, a theme already raised by clinicians in the survey (Chapter 5.6.7).

By contrast, current assessment technologies, particularly 3D gait analysis laboratories and robotic rehabilitation systems, remain largely misaligned with the operational and economic realities of community stroke care. Capital costs

in the six-figure range, plus maintenance, space, and specialist staffing, render them ill-suited for frequent review outside tertiary centres (Baker, 2013; Mehrholz et al., 2017). Their scarcity also amplifies geographic inequities: patients in rural or deprived areas, already at higher risk of stroke and disability (Marmot, 2020), are the least likely to access these services (Chapter 5.5).

The designation of the system as “low-cost” is relative to existing gait analysis modalities rather than an absolute unit price. As outlined in Chapter 6, material and component selection was deliberately guided by accessibility, favouring commercially available sensors, microcontrollers, and textile-based electrodes over bespoke or laboratory-dependent systems. In contrast, conventional three-dimensional gait analysis requires substantial capital investment, specialised laboratory infrastructure, and trained personnel, which limits accessibility and increases overall service cost (Carse et al., 2013).

The delivery of such assessments is also time-intensive, often requiring extended setup, data collection, and post-processing, further increasing clinical burden (Kelly et al., 2016). Importantly, cost in this context extends beyond hardware to include setup time, staffing requirements, and integration into clinical workflows. By minimising these demands, the present system reduces total service burden and barriers to access. “Low-cost” should therefore be understood as a system-level characteristic, reflecting reduced financial and logistical constraints compared to current gold-standard approaches.

The prototype hardware cost can be reasonably estimated at approximately £150 per unit. This estimate reflects a lower-bound cost based on the use of commercially available components and simple fabrication methods, as outlined in Chapter 6. Within this total, the compression sock with integrated copper electrodes represents the least standardised and most variable element. At the prototype stage, this component is relatively costly due to its

bespoke nature, requiring manual integration of conductive materials, individual fitting considerations, and iterative fabrication. These factors introduce inefficiencies that inflate unit cost. However, this cost is not intrinsic to the technology itself but to its current stage of development; with design refinement, standardisation of electrode placement, and transition to textile manufacturing processes (e.g., automated knitting or printed conductive pathways), substantial cost reductions would be expected.

In contrast, the remaining components, including the IMU platform, EMG acquisition board, and force sensor, are already mass-produced, widely distributed, and supported by established supply chains. Their inclusion in the design was therefore intentional, reflecting a strategy to leverage existing, low-cost electronics to minimise both unit cost and procurement barriers. These components also benefit from ongoing reductions in price driven by the broader consumer electronics market, further supporting scalability. Importantly, the use of modular, off-the-shelf hardware also reduces replacement and maintenance costs, as individual components can be sourced and substituted without reliance on proprietary systems.

When considered in relation to conventional three-dimensional gait analysis, the difference in cost becomes more pronounced. A fully equipped gait laboratory typically requires substantial capital investment, often in the range of tens to hundreds of thousands of pounds for motion capture systems, force plates, and associated infrastructure, alongside ongoing costs related to maintenance, software licensing, and specialist staffing (Carse et al., 2013; Kelly et al., 2016). In addition, each assessment is time-intensive, requiring extended setup, calibration, and post-processing, which further increases per-patient cost and limits throughput.

Against this backdrop, a textile EMG/IMU device designed for medical-grade manufacture could shift the cost–benefit calculus by orders of magnitude,

provided issues of durability, infection control, and regulatory compliance are addressed (Sections 6.6–6.7). Such a shift is not hypothetical but grounded in the feasibility findings of Chapter 8, where stroke survivors tolerated the garment, and clinicians reported willingness to adopt it if data were interpretable and workflows remained streamlined.

As indicated by these findings on garment tolerance, the garment can be used for extended durations without adverse effects. However, its intended role within the context of the reported work is not continuous or long-term monitoring, but as a session-based assessment tool. The intended purpose is to collect targeted data during discrete clinical encounters to inform rehabilitation pathways and dosing of treatments. Longitudinal data collection and continuous wear are not supported within the current system design.

The workforce implications further reinforce this argument. Physiotherapists, occupational therapists, and orthotists are all designated shortage occupations in the UK (UK GOV, 2023), and Chapter 5 confirmed that 54% of clinicians cite staffing as a barrier to objective gait assessment. A wearable system capable of compressing complexity into a handful of validated, interpretable indicators could help to offset this shortage, reducing reliance on highly specialised gait analysts and enabling broader professional groups to make safe, evidence-informed adjustments.

However, the potential for efficiency gains must be balanced against the risk of hidden workload. As Greenhalgh et al. (2017) caution, digital tools can inadvertently increase clinician burden if interfaces are clunky or if data streams are poorly integrated into records. The usability-focused design iterations reported in Chapters 6 and 8, compression-integrated electrodes, simplified calibration, and clinician-led selection of output variables, were therefore not incidental but central to the health service value proposition.

Health economic modelling will eventually require formal cost-effectiveness analyses, using endpoints such as quality-adjusted life years (QALYs) and incremental cost-effectiveness ratios. Comparable evidence from related fields suggests that wearable monitoring can reduce reliance on specialist services and enable earlier discharge, leading to cost savings (Rogers et al., 2019). Yet such claims must be tested empirically in the stroke context, with prospective studies designed to capture both economic and patient-centred outcomes. The thesis findings provide a proof-of-concept basis for this trajectory: laboratory comparisons (Chapter 7) confirmed the plausibility of robust temporal markers, while feasibility work (Chapter 8) demonstrated patient tolerance and clinician enthusiasm. The logical next step is not to claim definitive cost-effectiveness but to argue that the preconditions for cost-effectiveness, acceptable performance, usability, and integration potential, have now been established.

Nonetheless, several caveats temper the cost–benefit argument. First, a narrow focus on cost risks obscuring equity and quality. If deployment clusters in better-resourced centres, or if patients with low digital literacy cannot benefit equally, then efficiency gains may deepen rather than reduce disparities (George et al., 2020). Second, metrics fixation is a danger: if wearable outputs are interpreted in isolation: say, stance symmetry without context, they may prompt inappropriate therapy adjustments. Third, algorithmic bias remains a possibility if analytic models are trained on relatively mobile survivors, producing over-confident outputs in severely impaired users. These limitations argue strongly for evaluation frameworks that integrate patient-centred outcomes (comfort, usability, participation) and equity audits alongside economic endpoints. Indeed, the thesis findings in Chapter 8 already illustrate this principle, showing that comfort and donning practicality were decisive factors in patient tolerance, issues as central to real-world cost–benefit as any reduction in falls or admissions.

Taken together, the findings and arguments presented across this thesis suggest that the cost–benefit case for wearable EMG/IMU systems is less about raw economics and more about systemic fit. The true value proposition lies in shifting the geography and frequency of objective assessment, from scarce, centralised labs to routine community and bedside contexts, while reducing reliance on scarce specialist labour. The evidence generated here shows that this is feasible, tolerated, and valued by clinicians, creating the conditions under which formal cost-effectiveness can and should be tested. Whether those conditions translate into measurable system savings will depend less on the sensors themselves than on the organisational choices that determine how they are deployed, supported, and sustained.

9.8 Clinical and Theoretical Implications

9.8.1 Implications for Neuro-rehabilitation Theory and Practice

The findings presented across this thesis compel a rethinking of how gait is assessed and theorised within post-stroke rehabilitation. Historically, models of gait recovery have been grounded in laboratory-based constructs, with kinematic and kinetic measures derived from three-dimensional motion analysis used to infer deficits and prescribe interventions (Baker, 2013). While scientifically rigorous, such approaches privilege experimental control over accessibility, restricting their utility to a minority of stroke survivors who can tolerate and access laboratory conditions. The empirical evidence generated in Chapters 7 and 8 challenges this paradigm by demonstrating that meaningful activation–timing data can be captured using fabric-integrated EMG and shank-mounted IMUs in contexts far removed from the laboratory. This shift is not simply technological but conceptual: it positions gait recovery as something that can be measured and understood within the environments

where rehabilitation actually occurs, not just under idealised laboratory conditions.

The integration of EMG and IMU signals offers a particularly strong contribution to theoretical debates about motor control and recovery. The feasibility study demonstrated that, despite signal variability inherent to textile electrodes, it was possible to consistently identify activation–timing markers such as delayed tibialis anterior onset during late swing (Sections 7.6.3, 8.5). These findings align with contemporary theories suggesting that stroke recovery is mediated not by a simple restoration of pre-stroke motor patterns but through the reorganisation of muscle synergies and compensatory recruitment strategies (Cheung et al., 2009). In this sense, variability in EMG traces—often treated as an obstacle to signal processing—can itself be read as evidence of adaptive control processes, resonating with Stergiou and Decker’s (2011) argument that motor variability reflects functional flexibility rather than measurement “noise.”

Moreover, the reliability of inertial-derived temporal markers (stride time, cadence, and stance–swing phase distribution), which frequently achieve ICCs above .80 under controlled conditions (Mancini et al., 2011; Kobsar et al., 2020), suggests that portable systems can provide the scaffolding required to interpret noisy EMG data. The thesis findings therefore advance both theory and practice: they operationalise motor control frameworks in real-world contexts while offering clinicians interpretable temporal anchors against which muscle activity can be assessed. The clinical implication is a pragmatic middle ground between the unquantified subjectivity of observational gait analysis—still the norm for 66% of UK clinicians in the survey (Section 5.6.1)—and the inaccessible precision of 3D gait laboratories.

From a practical standpoint, this means that rehabilitation can be informed by short, objective captures taken at the bedside or during community reviews.

The feasibility study confirmed that such short bouts were well tolerated, with adjustable compression and coached set-up ensuring patient comfort (Section 8.5). This aligns with person-centred models of rehabilitation, which stress that assessment tools must not exacerbate fatigue, anxiety, or cognitive load in already vulnerable populations (Topol, 2019). It also directly addresses the appetite for innovation expressed in Chapter 5, where 82% of clinicians indicated willingness to adopt portable technologies if they were low-cost and interpretable. The wearable system developed here is therefore more than a technological prototype; it represents a conceptual advance toward embedding mechanistic insight into the everyday delivery of stroke rehabilitation.

9.8.2 Implications for Orthotic and Therapeutic Practice

If the theoretical implications concern how gait recovery is understood, the clinical implications are perhaps most acute in relation to orthotic and therapeutic practice. Orthotic provision—particularly the prescription and tuning of ankle–foot orthoses (AFOs)—was highlighted in the questionnaire study as an area of unmet need. Respondents noted that orthotic adjustments are often made based on visual estimates, without objective feedback, and that specialist gait laboratories are rarely accessible to validate these adjustments (Sections 5.6.3, 5.6.5). This thesis directly addresses that gap by demonstrating how wearable EMG/IMU data can provide interpretable indicators such as stance-time asymmetry and dorsiflexor activation windows that speak directly to orthotic function (Section 7.6.3).

The practical importance of this is twofold. First, it shifts orthotic practice from reactive to data-informed decision-making. Rather than waiting for visible signs of maladaptation—compensatory hip hiking, early heel rise, or patient-reported discomfort—clinicians could use wearable outputs to identify and correct gait

deviations early. Second, it enables more nuanced titration of orthoses over time.

As Chapter 2 argued, the effectiveness of AFOs is highly dependent on alignment tuning, particularly when considered through the shank-to-vertical angle (Bregman et al., 2011; Section 2.5.3). Yet in practice, tuning is often approximate, and re-assessment infrequent. By embedding objective temporal and activation data into routine reviews, this thesis offers a pathway to orthotic optimisation that is continuous, decentralised, and less dependent on scarce laboratory resources.

Therapists stand to benefit equally. The reliance on subjective judgement for therapy progression—a point repeatedly raised in free-text survey responses (Section 5.6.7)—means that treatment intensity and discharge readiness are often determined without precise biomechanical evidence. By providing quantifiable markers that are clinically interpretable (e.g., cadence stability, TA onset, symmetry ratios), wearable systems could enhance confidence in progression decisions and support more targeted therapy dosing. This, in turn, speaks to broader service-delivery models. As argued in Chapter 9.7, stroke rehabilitation is constrained by both workforce shortages and geographic inequity. A wearable system capable of producing objective, interpretable outputs in community and outpatient contexts represents a practical response to both challenges, enabling more patients to access quantifiable assessment without requiring specialist centres.

The implications also extend to theoretical debates about neuroplasticity and orthotic use. Critics have argued that AFOs may inhibit recovery by reducing the need for active dorsiflexor recruitment, potentially limiting neuroplastic adaptation (Tyson & Kent, 2013). The wearable system described here provides, for the first time, a tool capable of empirically interrogating this debate in everyday clinical contexts: does AFO use reduce TA activation in

swing, or does it enable survivors to practise safer, higher-dose walking that indirectly promotes recovery?

The answer to this question cannot be determined by subjective observation alone; it requires precisely the kind of temporal and activation-level monitoring demonstrated in this thesis. In this sense, the clinical and theoretical implications converge: the technology is not merely an adjunct to practice but a means of resolving contested debates about recovery versus compensation in post-stroke gait.

Taken together, these findings underscore that the clinical and theoretical contributions of the thesis are mutually reinforcing. By demonstrating that wearable systems can produce interpretable activation–timing data in real-world contexts, the research challenges laboratory-dominant paradigms of gait assessment while simultaneously addressing pressing clinical needs in orthotic tuning and therapy progression. The implications are therefore twofold: a theoretical reframing of how stroke recovery is understood, and a practical blueprint for integrating objective, low-burden assessment into the everyday practice of stroke rehabilitation.

9.8.3 Mapping Implications Back to the Central Research Question

The central research question addressed whether current and emerging approaches to gait assessment can meaningfully distinguish true neuromechanical recovery from compensatory adaptations. Evidence reviewed in Chapter 2 and reinforced by the survey findings in Chapter 5 demonstrates that conventional practice struggles with this distinction. Observational gait analysis—still the dominant tool in UK stroke services—cannot reliably differentiate, for instance, between improvements in walking speed achieved through compensatory hip circumduction and those reflecting restored plantarflexor propulsion (Sections 2.3.3.4, 5.6.1). Similarly, functional

outcome measures such as the 10MWT, while pragmatic and entrenched in clinical pathways, collapse performance into a single scalar value that obscures the underlying mechanisms driving change (Section 2.3.3.1).

The wearable EMG/IMU system developed and evaluated here was designed to address precisely this gap. By combining activation–timing data with inertial signals aligned to gait events, the system enables clinicians and researchers to interrogate whether functional improvements reflect neuromuscular reorganisation or reliance on energy-inefficient compensations. The feasibility work demonstrated that clinically relevant markers—such as stance-time asymmetry and delayed tibialis anterior onset—could be detected outside of laboratory infrastructure, underlining the potential for more nuanced monitoring in routine settings (Sections 7.6.3, 8.5). In this way, the system offers a mechanism for operationalising the compensation–recovery distinction within accessible clinical practice.

These implications extend to orthotic prescription, an area where mechanistic evaluation has historically been underdeveloped. Many intervention studies exclude AFO users, which restricts understanding of how orthotic tuning influences activation timing, propulsion, and overall recovery trajectories (Burpee & Lewek, 2015; Section 2.4.2). Wearable monitoring systems create the conditions for directly examining these interactions in clinical populations, allowing orthoses to be understood not merely as static supports but as dynamic modulators of motor control.

Taken together, these insights point toward a broader methodological shift. The future of stroke rehabilitation measurement need not involve a choice between the fidelity of laboratory analysis and the feasibility of clinical practice. Instead, the key lies in reconciling these approaches through wearable systems that capture critical mechanistic markers—timing, coordination, propulsion—in ecologically valid contexts. Such an approach provides the

tools necessary to answer the long-standing question at the heart of post-stroke gait rehabilitation: whether observed improvements signify recovery, compensation, or a complex interaction of both.

9.9 Limitations

All research is necessarily bounded by methodological, technical, and systemic constraints, and acknowledging these boundaries is essential for a rigorous interpretation of findings. The body of work presented here is no exception. Although the studies advance understanding of accessibility gaps in gait analysis and explore the feasibility of wearable EMG/IMU technologies, several limitations warrant critical consideration.

For example, the feasibility nature of Chapter 8 imposes clear limitations. Small subgroup sizes, especially for stroke+AFO participants, limit statistical generalisability and risk masking subtle between-group differences. Normalisation to non-stroke MVC preserved amplitude differences but may have exaggerated inter-individual variability. Furthermore, VAS provided only a unidimensional view of comfort and may underestimate the complexity of user experience. However, these limitations should not be viewed as methodological flaws but as inherent to early-phase feasibility work. The key outcome is the demonstration that such data can be collected safely and meaningfully, justifying progression to larger, more powered studies.

9.9.1 Technical, Methodological, and Systemic Constraints

The technical limitations of the prototype device form the most immediate boundary of interpretation. Fabric-integrated electrodes provided a low-burden, tolerable alternative to gel electrodes, but at the cost of stability and fidelity. Signal quality was highly sensitive to compression pressure and

garment fit, leading to fluctuations in amplitude and vulnerability to motion artefact—issues well documented in the wearable EMG literature (Merletti & Farina, 2016; Ankhili et al., 2019). While timing indices such as tibialis anterior onset relative to gait cycle were preserved with reasonable robustness (cf. Section 7.6.3), the amplitude-dependent features remained less reliable than those derived from laboratory-grade systems. This trade-off underscores that the current device should be considered a proof-of-concept system rather than a clinically deployable solution, consistent with early-phase technology development models (MRC, 2006).

The IMU component introduced its own compromises. Although inertial timing of cadence and stride length generally performs within acceptable limits of agreement ($ICC > .80$, $TE < 5\%$; Mancini et al., 2011; Kobsar et al., 2020), our Arduino-class hardware constrained sampling frequency, processing power, and durability. Drift and sensor slippage were observed, especially in longer trials, limiting the confidence with which continuous kinematic trajectories could be derived. These are not trivial issues: clinicians emphasised in the questionnaire (Section 5.6.7) the need for rapid, reliable outputs, and hardware instability risks undermining precisely the confidence that would encourage clinical adoption.

Methodological design choices further constrain interpretation. The national questionnaire study offered valuable insights into UK clinical practice but achieved only a modest response rate, raising the risk of response bias—particularly an overrepresentation of clinicians with a pre-existing interest in gait assessment (Section 5.4). Similarly, the laboratory validation and feasibility cohorts were necessarily small, reflecting common recruitment difficulties in stroke rehabilitation research (Eldridge et al., 2016). Stroke survivors frequently present with fatigue, reduced mobility, or cognitive load that makes participation in repeated walking trials demanding, and these very factors may have skewed the sample toward higher-functioning individuals.

While small-sample feasibility designs are appropriate at this developmental stage, the diversity of stroke presentations—chronicity, severity, comorbidities—means that generalisability across the wider stroke population must be treated with caution (Winstein et al., 2016).

A specific limitation of the feasibility study detailed in Chapter 8 relates to the level of contextual clinical information reported for Stroke-affected participants prior to intervention. Although all participants underwent comprehensive clinical assessment to determine safety, suitability, and appropriateness for inclusion within the study, detailed reporting of baseline clinical presentation and pre-intervention gait characteristics was limited within the final study documentation. Inclusion of more extensive participant profiling, including clinical gait presentation, orthotic history, functional status, and biomechanical characteristics, would have provided greater contextual depth and improved interpretation of the findings.

This reflects increasing recognition within stroke orthotic rehabilitation literature that interpretation of gait and orthotic outcomes requires sufficient contextual clinical information, including baseline gait presentation, functional status, and multidisciplinary clinical assessment, to support meaningful translation into practice (Golding-Day et al., 2024). Future studies must therefore place greater emphasis on comprehensive participant characterisation and baseline gait reporting alongside primary outcome measures.

A further limitation lies in the tension between ecological validity and methodological control. Laboratory validation provided the fidelity necessary for benchmarking against gold standards, but in doing so stripped away many of the contextual challenges—donning variability, perspiration artefact, fatigue—that emerged in the feasibility study (Section 8.6). Conversely, the feasibility trial captured these real-world frictions but at the expense of

controlled precision. This trade-off between internal and external validity is not unique to this project; it reflects a longstanding challenge in rehabilitation engineering where devices that succeed under controlled conditions often falter when exposed to the contingencies of routine practice (Patel et al., 2012).

Interpretability of outputs represents another important limitation. At this stage of development, substantial researcher input was required to process raw EMG and IMU signals, apply filters, and extract clinically interpretable metrics. Manual correction was often necessary to stabilise event detection. While this approach sufficed in a research environment, it highlights the gap between proof-of-concept feasibility and clinical applicability.

Clinicians consistently articulated the need for “few-click,” one-page summaries with clear indicators and uncertainty bounds (Section 5.6.8). Without streamlined software pipelines, automated analytics, and user-friendly interfaces, wearable systems risk overwhelming rather than supporting clinical decision-making.

Finally, even assuming technical and methodological issues are resolved, systemic barriers to adoption remain profound. The NHS workforce faces well-documented shortages in physiotherapy, occupational therapy, and orthotics (UK GOV, 2023). Any technology that increases workload, requires specialist interpretation, or adds friction to existing workflows risks compounding rather than alleviating pressures.

The adoption of wearable systems thus cannot be reduced to hardware performance alone: it is fundamentally shaped by organisational structures, training pathways, and integration with health information systems. Technologies that fail to align with these realities may reinforce inequities rather than resolve them.

9.9.2 COVID-19 Constraints

An additional layer of limitation arose from the COVID-19 pandemic, which coincided with much of the data collection phase. National lockdowns and institutional restrictions imposed at the University of Strathclyde during 2020–2021 profoundly reshaped the scope and conduct of the research. Recruitment of stroke survivors was severely curtailed, both because of institutional policies restricting access to vulnerable participants and because of widespread hesitancy among patients to attend in-person sessions. Laboratory access was rationed, protocols for cleaning and PPE slowed throughput, and only staff and students with campus clearance could participate.

These restrictions necessitated compromises in study design. The proof-of-concept work (Chapter 7) relied heavily on healthy volunteers rather than stroke survivors, limiting the immediate generalisability of validation findings to clinical populations. The feasibility study (Chapter 8), while eventually including stroke participants, was constrained in size and required shortened protocols to minimise fatigue and infection risk. The net effect was a narrowing of the empirical base, both in terms of participant diversity and data volume. Moreover, participants who did engage were likely skewed toward those who were younger, more mobile, and less clinically vulnerable, potentially biasing feasibility findings toward a more optimistic view of device tolerance and usability.

These constraints are not idiosyncratic but reflect a global challenge for rehabilitation research during the pandemic, when in-person studies with clinically vulnerable groups were widely delayed, curtailed, or suspended (Watermeyer et al., 2021). From a methodological perspective, this context should temper interpretation of findings: what was feasible under COVID-19 conditions may not fully represent what is feasible under standard service delivery. From a systemic perspective, however, the pandemic also

accentuated the salience of portable, decentralised technologies—precisely because they reduce reliance on tertiary centres and extended in-person sessions. In this sense, the COVID-19 context, while restrictive, also underscored the urgency of developing pragmatic, low-burden solutions for stroke rehabilitation.

9.10 Future Possibilities

The findings of this research provide a foundation, but they also expose the methodological, technical, and systemic challenges that must be addressed before wearable EMG/IMU systems can become embedded in stroke rehabilitation practice. Future directions should not be interpreted as generic opportunities for further study; they represent specific, evidence-driven priorities shaped by the empirical results presented in Chapters 5–8 and by the inequities and gaps in current practice identified throughout the thesis. These priorities can be grouped into three interrelated domains: technical and analytic refinement, clinical integration with orthotic and therapeutic practice, and organisational adoption and workforce development.

9.10.1 Technical and Analytic Refinement

The technical limitations identified in Chapter 7 and 8 underline the need for focused refinement of both hardware and analytics. Textile electrodes integrated into compression garments demonstrated feasibility and tolerability but also introduced variability in skin contact, leading to fluctuations in signal quality. Addressing this limitation will require collaboration between rehabilitation scientists and textile engineers to optimise electrode geometry, material properties, and compression stability.

Evidence from textile EMG literature shows that electrode pressure in the range of 15–20 mmHg is optimal for minimising impedance and motion artefact (Ankhili et al., 2019) yet achieving this consistently across diverse patient anatomies remains a challenge. The next stage of development must therefore involve iterative testing with larger, more varied stroke populations to identify thresholds where clinical usability and signal fidelity can coexist.

Equally important is the progression beyond Arduino-class platforms. While these microcontrollers facilitated rapid prototyping, their computational constraints limited real-time processing and robustness. Transitioning to medical-grade electronics with higher sampling rates and embedded pre-processing is essential for clinical translation. This will also enable compliance with regulatory standards such as UKCA/CE marking, which demand not only technical safety but also cyber-security and data-governance assurances. Without such transition, devices risk remaining confined to academic settings, never reaching the clinical populations that survey respondents in Chapter 5 indicated were most in need of objective assessment tools.

Beyond hardware, analytic refinement is critical. The current system required researcher-led processing of raw signals, a bottleneck incompatible with NHS clinical workflows. Automated algorithms must be developed that can reliably detect gait events, quantify asymmetry, and highlight phase-specific activation changes. Crucially, these algorithms must account for the variability inherent in post-stroke motor control, distinguishing adaptive variability from maladaptive compensations (Stergiou & Decker, 2011). This thesis provided preliminary evidence that pattern-level features—such as tibialis anterior onset relative to swing—are robust and interpretable (Section 7.6.3). Future analytic pipelines should therefore prioritise such features while explicitly flagging uncertainty, avoiding overconfidence in marginal data. This refinement would make outputs both more trustworthy and more directly actionable for clinicians.

9.10.2 Clinical Integration with Orthotic and Therapeutic Practice

A recurring theme in both the questionnaire and feasibility work (Chapters 5 and 8) was the unmet need for objective data to guide orthotic prescription and tuning. Clinicians reported that orthotic decisions are often based on visual estimation, with limited access to mechanistic information about how ankle–foot orthoses influence gait dynamics. This gap is particularly problematic given the sensitivity of outcomes to shank-to-vertical angle tuning (Kerr et al., 2020) and the risk that poorly matched devices reinforce compensatory patterns (Burpee & Lewek, 2015). By offering quantifiable markers of stance-time asymmetry and phase-specific activation timing, wearable EMG/IMU systems could provide precisely the information that is currently absent from orthotic decision-making.

Future research should therefore embed wearable assessments directly into orthotic pathways, testing whether real-time or short-bout feedback improves the precision of AFO tuning and reduces compensatory strategies. Such integration would respond directly to clinicians stated needs (Section 5.6.3) and move the field toward combined therapy-plus-orthosis protocols, which remain underexplored despite being advocated in the literature (Jonkers et al., 2009). In this way, wearables would not replace orthoses but serve as an adjunct, enabling their optimal deployment.

Therapeutically, the capacity to monitor activation–timing patterns in routine clinical or community settings would support more targeted progression of therapy intensity. For example, the ability to document improvements in trailing limb angle or gastrocnemius recruitment outside the laboratory could guide therapists in adjusting training load or selecting adjunctive modalities such as FES. Importantly, such integration moves assessment from a reactive model, identifying deficits after they manifest, to a proactive model in which subtle maladaptation is identified and addressed early. The thesis has already

demonstrated proof of concept for short, tolerable assessments in stroke survivors (Section 8.5); the next step is to evaluate whether incorporating such assessments into therapy schedules alters outcomes in gait speed, endurance, and participation.

9.10.3 Organisational Adoption and Workforce Development

Even with technical improvements and clinical applications, wearable technologies will fail if they do not align with organisational realities. The NHS context, as documented in Chapters 1 and 5, is characterised by workforce shortages across physiotherapy, occupational therapy, and orthotics (UK GOV, 2023). Introducing devices that require specialist interpretation or lengthy training risks exacerbating rather than alleviating pressures. The survey (Section 5.6.7) was unequivocal: clinicians want tools that are fast, interpretable, and integrated with existing workflows.

Future directions must therefore include rigorous pilot implementation studies that evaluate not only device accuracy but also workflow impact, time-to-completion, and clinician confidence. Training packages need to be concise, cross-disciplinary, and scalable, designed so that outputs can be interpreted by multiple professional groups. Integration with electronic health records and national audit frameworks such as SSNAP would also be essential, ensuring that wearable-derived metrics are recognised within performance monitoring and quality assurance. Without such system-level integration, even technically successful devices risk becoming isolated “add-ons” that fail to shift practice at scale.

Policy-level engagement will also be critical. As seen during the COVID-19 pandemic, where remote and hybrid models of rehabilitation were rapidly implemented (Section 9.9.6), health services are capable of transformative change when necessity dictates. Wearable systems are well positioned to

support the current policy emphasis on decentralised, home-based care (NHS England, 2019), but evidence of scalability, cost-effectiveness, and equity will be necessary to secure commissioning. Importantly, deployment must guard against digital exclusion, ensuring that patients with lower health literacy or limited digital access are not left behind (George et al., 2020). Evaluation of equity impacts must therefore be embedded into all future implementation trials.

9.11 Summary

The work undertaken demonstrates that the barriers to equitable gait assessment in stroke rehabilitation are not solely technical, but systemic, methodological, and conceptual. Across the preceding chapters, it was shown that access to high-fidelity gait analysis remains severely limited (Sections 5.5–5.6, 9.4), that orthotic assessment is too often guided by observation rather than quantifiable parameters (Sections 2.5, 5.6.3), and that workforce shortages constrain the extent to which even available technologies can be deployed (Sections 5.6.5–5.6.7). Against this backdrop, the studies presented here argue for an alternative model: short, objective, and portable assessments that complement rather than compete with laboratory systems and that expand the reach of biomechanical insight into everyday practice.

The technical contribution of the fabric-based EMG/IMU system illustrates that clinically meaningful information can be captured without requiring full laboratory infrastructure. While the fidelity of EMG amplitudes remains limited compared with gel electrodes (Merletti & Farina, 2016), the combination of activation timing with inertial markers provides consistent pattern-level data—precisely the type of information most relevant for clinical interpretation (Sections 7.6.3, 8.6). This represents not just an engineering innovation but a

reframing of what counts as “good enough” evidence for guiding therapy: accuracy need not equal laboratory precision if the indicators are interpretable, repeatable, and capable of changing decisions in practice.

Clinically, the results speak directly to the inequities identified in the national questionnaire survey. With 66% of respondents relying on observation alone and only 3% reporting access to gait laboratories (Section 5.6.1), the appetite for accessible, objective tools was evident. The wearable prototype does not replace expert analysis, but it offers a tiered alternative: objective indicators at the bedside or community clinic, reserving laboratory capacity for complex cases. This redistribution of assessment capacity aligns with broader health system imperatives to decentralise care and to embed digital tools that reduce rather than intensify inequity (Winstein et al., 2016; NHS England, 2019). Importantly, the feasibility study confirmed that stroke survivors tolerated short, coached bouts with minimal discomfort (Section 8.5), underscoring that usability can be achieved without sacrificing patient experience.

At a theoretical level, the integration of EMG and IMU data provides a means of operationalising contemporary models of motor recovery. The feasibility findings: delayed tibialis anterior onset in late swing, persistent stance asymmetry, support the proposition that post-stroke walking is shaped less by restoration of pre-morbid motor patterns than by reorganisation of synergies (Cheung et al., 2009; Section 2.3.3). By embedding these observations in ecologically valid contexts, the work advances a shift in rehabilitation science from laboratory-defined constructs toward real-world motor control. Such a shift is critical if theory is to remain relevant to the environments in which recovery unfolds.

The health-economic implications are equally compelling. Stroke costs the UK an estimated £26 billion annually, with long-term disability accounting for the majority of this burden (Stroke Association, 2023; Section 1.3). Even modest

improvements in orthotic tuning or therapy targeting could reduce secondary complications, falls, or unplanned admissions. Wearable-assisted assessments, if integrated into structured review points, have the potential to deliver these gains at scale by providing earlier and more precise feedback loops (Sections 9.7.3–9.7.5). However, as argued throughout, these benefits are contingent not only on technical refinement but also on organisational will: devices must be embedded into workflows, supported with training, and aligned with service priorities, or else risk contributing to hidden workload and digital inequity (Sections 9.4.4, 9.7.6).

Taken together, the contribution here lies not in offering a finished device, but in reframing how stroke rehabilitation might balance precision with accessibility. By demonstrating that meaningful gait data can be collected through wearable, low-cost systems, and by situating this within the realities of clinical practice and health service delivery, the work opens a pathway for more equitable and personalised models of care. The argument is not to displace laboratories or high-cost technologies, but to extend the reach of quantifiable assessment into the spaces where recovery is lived—community clinics, outpatient reviews, and the home. In this way, the evidence presented not only advances technical innovation but also offers a reorientation of rehabilitation science toward scalable, patient-centred solutions.

Chapter 10 – Conclusion and Impact

10.1 Achievement of Research Aims

This thesis set out three aims (Chapter 3): (1) to characterise UK stroke-rehabilitation gait assessment practice and access; (2) to design and prototype an orthosis-compatible, wearable EMG–IMU system that prioritises short set-up and interpretable outputs; and (3) to validate and feasibility-test the system from controlled to real-world contexts so that its role in service pathways can be specified.

Aim 1 is supported. The national survey (Chapter 5) demonstrated a high, persistent reliance on observational gait analysis with constrained access to instrumented methods and laboratories (see Sections 5.5–5.6). Respondents indicated clear appetite for objective, portable tools provided they are quick to apply and clinically interpretable (Section 5.6.7–5.6.8). These findings align with rehabilitation guidance prioritising task-specific, dose-intense, and measurable interventions (Winstein et al., 2016) and substantiate the need case set out in Chapter 1.

Aim 2 is supported. Through iterative, action-research–style prototyping (Chapter 6), the thesis delivered a textile EMG garment integrated with an IMU and modular electronics, designed for orthosis compatibility, rapid donning, and repeatable electrode placement (see Sections 6.3–6.7). The design requirements were explicitly derived from end-user constraints collected in Chapter 5 and from orthotic-tuning use-cases reviewed in Chapter 2.

Aim 3 is supported with clearly defined boundaries. Laboratory comparisons (Chapters 7 and 8) indicate acceptable agreement for temporal spatiotemporal parameters relative to reference systems (e.g., cadence/stride-

time behaviour consistent with ICCs typically $> .80$ and typical errors $< 5\%$ reported in the literature), while amplitude-heavy EMG metrics remain more variable with textile electrodes (Mancini et al., 2011; Merletti & Farina, 2016; Kobsar et al., 2020; see Sections 7.4–7.6, esp. 7.6.3). Feasibility with stroke survivors (Chapter 8) demonstrated tolerability and operational feasibility during short, coached bouts, and surfaced ecological constraints (perspiration artefact, micro-slippage, donning variability) that now define the engineering and analytics roadmap (Sections 8.4–8.6; 9.10.1–9.10.2). In translational terms, the system is fit-for-purpose for timing-dominant indicators outside the lab, with a clear plan to harden amplitude-sensitive measures.

10.1.1 Summary of thesis outcomes (what was concretely delivered)

A needs and access map for UK stroke gait assessment identifying geographic, procedural, and workforce barriers and a stated clinician appetite for portable, interpretable tools (Chapter 5; Sections 5.5–5.6).

A clinically wearable, orthosis-compatible textile EMG and IMU prototype with compression zoning and repeatable electrode geometry, engineered to minimise set-up time (Chapter 6; Sections 6.3–6.7).

Bench/lab validation demonstrating robustness of event timing and cadence/stride-time estimation relative to reference systems and specifying the limits of amplitude fidelity with textiles (Chapter 7; Sections 7.4–7.6).

Feasibility evidence in stroke survivors showing tolerability, recoverable data rates during short bouts, and the operational issues that matter in routine use (Chapter 8; Sections 8.4–8.6).

A service-facing use case and SOP-style recommendations positioning the device as a complement to 3D gait labs, not a replacement, with proposed

clinician-facing summary indicators (stance-time asymmetry, cadence banding, phase-specific activation timing) (Chapter 9; Sections 9.5–9.8, 9.10).

10.1.2 Summary of thesis conclusions (what follows from the evidence)

In services where labs are scarce, short, timing-dominant objective captures are both useful and realistic to deploy between infrequent lab visits (Chapters 5, 7–8; 9.5–9.6).

Textile EMG can deliver phase-appropriate activation timing through an AFO when paired with IMU-derived gait events, provided donning and compression are standardised (Chapters 6–7).

The value proposition is clinical interpretability and frequency of review, not decimal-place replication of lab amplitudes; this reframing is consistent with pragmatic, translational evaluation of rehabilitation technologies (Craig et al., 2008; Patel et al., 2012) (Chapter 9; Sections 9.3.3, 9.5.2).

Equity can be advanced when objective assessment moves closer to the point of care and into community reviews, if deployment includes training, simple outputs, and governance (Chapter 5; Sections 9.4, 9.7).

10.2 Justification of Novelty

Novelty lies not simply in the technical integration of EMG and IMU but in proving their feasibility in clinical populations. Chapter 8 demonstrated that meaningful signals could be obtained even in stroke survivors with spasticity, and that devices could be deployed safely in real-time gait testing. Few studies have combined neuromuscular, biomechanical, functional, and subjective

outcomes in a single framework, and the Chapter 8 findings highlight both the promise and the design limitations of such an approach.

10.2.1 Orthosis-compatible textile EMG integrated with inertial sensing in a single garment

The thesis demonstrates in-brace, phase-appropriate activation timing via a textile EMG garment fused with IMU event detection, engineered for rapid, repeatable set-up and compatible with common AFOs (Chapters 6, 7 and 8). This addresses a specific gap identified in the review (Chapter 2) where in-clinic EMG under orthoses is rarely feasible (Merletti & Farina, 2016).

10.2.2 A pragmatic, timing-dominant metric set for routine decisions

Rather than exporting full inverse dynamics to the clinic, the work prioritises trendable indicators, stance-time asymmetry, cadence banding, and event-linked activation timing, selected for clinical decision utility and robustness across contexts (Muro-de-la-Herran et al., 2014; Kobsar et al., 2020) (Sections 7.6.3; 9.5.2).

10.2.3 A service-centric action-research pipeline linking inequity to engineering

The device requirements were derived from service constraints (access, workforce, hygiene) and iterated with end-users, explicitly targeting decentralised capture (Chapters 5–6; Section 6.6; 9.4). This is a distinctive translational stance compared with technology-first approaches (Craig et al., 2008).

10.2.4 First steps toward an orthotic-aware evidence base

By capturing activation timing and gait events with the orthosis in situ, the thesis opens a route to data-informed orthotic tuning and follow-up that typical lab-only paradigms cannot deliver at the required frequency (Sections 7.6.3; 9.8.3).

10.3 Impact Statement

10.3.1 Clinical practice and pathways

The immediate impact is a low-friction adjunct to routine reviews that can be deployed at bedside, in outpatient clinics, or on community visits in ≤ 10 minutes, generating outputs of stance-time asymmetry, cadence banding, and phase-specific activation timing aligned to gait events (Chapters 7 and 8). These data currently require interpretation and analysis but it is predicted that future iterations of the design will allow for a simpler, single-page summary output. This supports orthotic tuning (e.g., dorsiflexion assist relative to terminal swing) and therapy progression decisions that currently rely on observation (Sections 5.6.5–5.6.8; 9.8.2–9.8.3).

However, the findings indicate that while devices may normalise timing, they do not address amplitude deficits; clinical pathways should therefore combine orthotic provision with targeted strengthening. Moreover, patient comfort must be addressed if devices are to be adopted in routine practice.

10.3.2 Patients and equity

Shifting objective capture closer to the point of care widens access for those with mobility and transport barriers (Section 9.4.1–9.4.4). The short, coached capture aligns with person-centred principles and tolerability demonstrated in feasibility (Section 8.5), and with policy ambitions for technology-enabled,

community-anchored rehabilitation (NHS England, 2019). It must be remembered however that without addressing comfort (as seen in Chapter 8 VAS scoring), patients may reject devices despite biomechanical benefits. Equity demands that supportive technologies are not only effective but also acceptable across diverse patient groups.

10.3.3 Research and innovation

The work defines testable thresholds for multicentre trials: a priori accuracy for event timing; test–retest reliability; minimal detectable change for stance asymmetry; and responsiveness to orthotic/therapy adjustments (Sections 7.6.3; 9.10.3). It also motivates analytics that link coordination quality (activation timing, synergy structure) to function, using automated pipelines for clinician-facing summaries (Section 9.10.2) and non-linear variability where appropriate (Stergiou & Decker, 2011).

10.3.4 Limitations and boundaries of inference

Textile EMG amplitude remains more variable than gel; timing is the robust signal outside the lab (Merletti & Farina, 2016). IMU precision is speed-dependent and sampling-limited; amplitude-heavy biomechanical inferences should be reserved for lab settings (Sections 7.6, 9.9.1). Small feasibility cohorts limit subgroup inference but were appropriate to the development stage (Eldridge et al., 2016) (Section 9.9.2).

10.3.5 Recommendations and future work

Future work should prioritise both engineering refinement and clinical translation. On the engineering side, specific developments are required to enhance data quality and device usability. Compression mapping and

electrode geometry should be optimised to ensure stable skin contact and reproducible EMG acquisition, while artefact-aware filtering and medical-grade microcontroller units (MCUs) are necessary to reduce signal noise and ensure robustness across diverse patient presentations (see Section 9.10.1). These refinements would address some of the technical barriers identified in Chapter 8 and align the platform with regulatory expectations for medical-grade monitoring systems.

Automation represents the second priority. Reliable real-time gait-event detection, accompanied by uncertainty bands, is critical to reduce analyst burden and support clinician-facing decision aids. A concise, one-page clinical output (Section 9.10.2) could bridge the gap between complex waveform analytics and day-to-day rehabilitation practice, ensuring that outputs are interpretable and actionable without requiring advanced technical expertise.

Third, validation must progress beyond single-centre feasibility to multicentre studies with predefined thresholds for acceptable accuracy and reliability (Section 9.10.3). Larger, heterogeneous samples will provide the statistical power needed to evaluate both between-group effects and within-group variability, while also testing the generalisability of findings across clinical settings.

Finally, the implementation pathway must include structured evaluation of economic viability. Budget-impact analyses and cost-effectiveness modelling (Section 9.10.5) will be essential to demonstrate value within resource-constrained healthcare systems. Such analyses should explicitly integrate an equity perspective (Section 9.4.4), ensuring that device solutions do not exacerbate disparities in access to care but instead provide affordable, scalable tools for rehabilitation across diverse populations.

Taken together, these priorities establish a roadmap from feasibility to adoption: engineering refinements to ensure technical reliability, automation to enable clinical usability, multicentre validation to demonstrate robustness, and economic evaluation to justify implementation. Each step is necessary to move beyond proof-of-concept and deliver a clinically relevant, equitable, and sustainable gait assessment solution.

10.4 Final remarks

The core contribution of this work is intentionally complementary to existing gait laboratory methods: rather than supplanting gold-standard settings, it seeks to extend access, continuity, and clinical utility. The developed platform enables more frequent, equitable, and objective insight into gait deviations between sparse laboratory assessments; translates complex multi-domain data into accessible indicators that clinicians can interpret rapidly; and embeds gait-assessment capability at the point of care. In doing so, the work proposes a scalable trajectory toward more timely, equitable, and data-informed stroke rehabilitation (Craig et al., 2008; Patel et al., 2012; Winstein et al., 2016).

Crucially, this aligns with the strategic priorities articulated in the NHS Long Term Plan (NHS England, 2019) and the NHS 10-Year Innovation and Technology Roadmap (NHS England, 2023), both of which call for expanded use of digital technologies to deliver *personalised, preventative, and community-based rehabilitation*. These policy frameworks emphasise reducing health inequities through technology-enabled monitoring and tailoring interventions to individual recovery profiles—precisely the aims operationalised by the present system. By supporting granular, real-time feedback on gait function, such tools can inform adaptive orthotic tuning and bespoke therapy progression, directly contributing to the NHS goal of data-driven, patient-centred care pathways.

It is essential for the reader to appreciate that the value of this research lies not merely in novel sensor hardware or bespoke signal pipelines, but in the integration of those advances into the clinical decision-making process—balancing analytic depth, operational simplicity, and professional acceptability. A device that is biomechanically superior but clinically unusable would fail its purpose. The feasibility findings in Chapter 8 illustrated this tension clearly: objective accuracy was achieved, yet practical constraints, user tolerability, and limited statistical power underscore the real-world challenges that must be overcome for clinical translation.

Thus, the final claim is a deliberate one: this work does not conclude the search for definitive orthotic or rehabilitation solutions but rather opens a pathway toward a sustainable, clinician-centred paradigm in gait monitoring—one that resonates with current NHS policy ambitions for personalised, technology-supported rehabilitation. Future work must advance that pathway through iterative design, validation across broader and more diverse populations, and ensuring that what is technically possible becomes clinically adoptable and systemically scalable within the evolving landscape of UK healthcare.

**“There’s no more work. We’re
destitute. I’m afraid I have no
choice but to sell you all for
scientific experiments.”**

– *Monty Python*

“The Meaning of life”

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Appendix A – Individual Database Search terms

PubMed/MEDLINE

Approach: MeSH terms (exploded where appropriate) combined with free-text in title/abstract. Field tags included [MeSH Terms], [Title/Abstract] ([tiab]), and [All Fields].

((("Stroke"[MeSH] OR stroke[tiab] OR "cerebrovascular accident"[tiab] OR cva[tiab])*

AND ("Gait"[MeSH] OR gait[tiab] OR walk*[tiab] OR "gait rehabilit*[tiab])*

AND ("Electromyography"[MeSH] OR electromyograph[tiab] OR EMG[tiab]))*

NOT ("Upper Extremity"[MeSH] OR "upper limb"[tiab] OR "upper extremity"[tiab] OR arm[tiab] OR hand[tiab] OR finger[tiab])*

AND ("2004/01/01"[Date - Publication] : "2024/12/31"[Date - Publication])

Targeted device layer (applied when required): AND ("ankle-foot orthos[tiab] OR AFO[tiab] OR orthos*[tiab] OR splint*[tiab]).*

EMBASE (Ovid)

Approach: Emtree terms with explode (exp) plus title/abstract/keyword free-text (.ti,ab,kw.). Proximity operators (adj#) bound concept pairs.

(exp cerebrovascular accident/ OR stroke.ti,ab,kw.)*

AND (exp gait/ OR gait.ti,ab,kw. OR walk*.ti,ab,kw. OR (gait adj3 rehabilit*).ti,ab,kw.)*

AND (exp electromyography/ OR electromyograph.ti,ab,kw. OR EMG.ti,ab,kw.)*

NOT (exp upper extremity/ OR arm.ti,ab,kw. OR hand*.ti,ab,kw. OR finger*.ti,ab,kw.)*

limit 2004-2024, humans, english, adult

CINAHL (EBSCOhost)

Approach: CINAHL Headings (MH, exploded with +) combined with free-text across TI/AB/TX. Proximity operators N# (any order) and W# (in order).

((MH "Stroke+") OR TI(stroke) OR AB(stroke*) OR TX("cerebrovascular accident" OR cva))*

AND ((MH "Gait+") OR TX(gait OR walk* OR "gait rehabilit*"))*

AND ((MH "Electromyography+") OR TX(electromyograph OR EMG))*

NOT ((MH "Upper Extremity+") OR TX("upper limb" OR "upper extremity" OR arm OR hand OR finger))*

Limiters: English; Peer Reviewed; Human; Adult; 2004–2024

Scopus (Elsevier)

Approach: TITLE-ABS-KEY field. Proximity operators W/# and PRE/# used to constrain phrase variants.

TITLE-ABS-KEY((stroke OR "cerebrovascular accident" OR cva)

AND (gait OR walk OR "gait W/2 rehabilit*"))*

AND (electromyograph OR emg))*

AND NOT TITLE-ABS-KEY("upper limb" OR "upper extremity" OR arm OR hand OR finger)*

AND PUBYEAR > 2004 AND PUBYEAR < 2024 AND (LIMIT-TO(LANGUAGE, "English"))

Web of Science Core Collection

Approach: Topic search (TS=) spanning title, abstract, author keywords, and Keywords Plus. Proximity operator NEAR/# imposed local context.

TS=((stroke OR "cerebrovascular accident" OR cva)

AND (gait OR walk OR "gait NEAR/2 rehabilit*"))*

AND (electromyograph OR EMG))*

NOT TS=("upper limb" OR "upper extremity" OR arm OR hand OR finger)*

Timespan: 2004–2024; Language: English; Document Types: Article, Review, Proceedings Paper

Cochrane Library

Approach: MeSH descriptors (exploded) combined with ti/ab/kw terms to identify trials and reviews relevant to gait and EMG after stroke.

[Stroke]:MeSH explode AND ([Gait]:MeSH explode OR gait:ti,ab,kw OR walk:ti,ab,kw)*

AND ([Electromyography]:MeSH explode OR electromyograph:ti,ab,kw OR EMG:ti,ab,kw)*

NOT ([Upper Extremity]:MeSH explode OR "upper limb":ti,ab,kw OR arm:ti,ab,kw OR hand:ti,ab,kw OR finger:ti,ab,kw)*

Limits: 2004–2024; English

ScienceDirect (supplementary, publisher platform)

Approach: Title/abstract/keyword search with platform filters for article type and subject area. Used as a supplementary source to surface in-press items and publisher-specific content.

TITLE-ABSTR-KEY(stroke OR "cerebrovascular accident" OR cva)

AND TITLE-ABSTR-KEY(gait OR walk OR "gait rehabilit*")*

AND TITLE-ABSTR-KEY(electromyograph OR emg)*

NOT TITLE-ABSTR-KEY("upper limb" OR "upper extremity" OR arm OR hand OR finger)*

Filters: Research/Review/Book chapter; 2004–2024; English; relevant subject areas

A.1 PICO Table of Literature Review Papers

Study ID (Authors, Year)	Population (P)	Intervention (I)	Comparator (C)	Primary Outcomes (O)	Key Findings	Notes
Adiputra, D., Nazmi, N., Bahiuddin, I., Ubaidillah, I., Imaduddin, F., Rahman, M. A. A., Mazlan, S. A., & Zamzuri, H. (2019)	Adults with gait impairment (including stroke)	AFO/orthotic	Controlled comparison	Stiffness modulation; damping; actuation; user adaptation	Emerging adjustable/active AFOs enable context-specific support; evidence for functional gains is promising but heterogeneous.	Narrative review
Alingh, J., Groen, B. E., van Asseldonk, E. H. F., Geurts, A. C. H., & Weerdesteyn, V. (2020)	Adults post-stroke	EMG/kinematic biofeedback; FES; Treadmill/velocity training	Not stated	Paretic propulsion metrics; walking speed; kinematics	Fast treadmill walking, FES to plantarflexors, and biofeedback show consistent improvements in paretic propulsion; evidence quality varies.	Systematic review
An, S., Lee, Y., Shin, H., & Lee, G. (2015)	Adults post-stroke (n=103)	Rehab/assessment (unspecified)	Controlled comparison	Gait speed; 6MWT distance; community walker classification	Gait speed (cut-off ≈ 0.87 m/s) better predicted community ambulation than 6MWT distance in moderately affected survivors.	Observational

Awad, L. N., Reisman, D. S., Kesar, T. M., & Binder-Macleod, S. A. (2014)	Chronic stroke (n=13; 12 completers)	FES	Not stated	Paretic propulsion, speed, balance, participation	Feasible/safe; improved paretic propulsion accompanied by functional gains maintained at 3 months.	Within- subject/Pre- post
Awad, L. N., Reisman, D. S., Kesar, T. M., & Binder-Macleod, S. A. (2016a)	Chronic stroke	FES; Treadmill/velocity training	Controlled comparison	Paretic propulsion; walking speed; participation	Targeting paretic propulsion using FastFES improved gait function versus comparison in a pilot RCT.	Randomised
Awad, L. N., Reisman, D. S., Pohlig, R. T., & Binder-Macleod, S. A. (2016b)	Chronic stroke	Instrumented gait analysis	Not stated	Propulsion, kinematics, clinical gait outcomes	Biomechanics-based grouping improved prediction of who benefits from targeted gait interventions.	Observational; Within- subject/Pre- post
Balasubramanian, C. K., Bowden, M. G., Neptune, R. R., & Kautz, S. A. (2007)	Adults post-stroke	Instrumented gait analysis	Not stated	Step length asymmetry; walking speed; energy cost	Greater step length asymmetry associated with worse walking performance.	Observational
Barroso, F. O., Torricelli, D., Moreno, J. C., Taylor, J., Gomez-	Adults post- stroke (n=9) vs healthy controls	EMG-based assessment	Not stated	Kinematics; kinetics; spatiotemporal parameters;	A multivariate set of kinematic, kinetic, temporal and synergy features (including paretic propulsion) predicted impaired	Observational

Soriano, J., Bravo-Esteban, E., Piazza, S., Santos, C., & Pons, J. L. (2017)				EMG synergies; paretic propulsion	walking better than Fugl–Meyer scores.	
Baudet, A., et al. (2014)	Adults (incl. post-stroke in application)	EMG-based assessment	Not stated	Surface EMG quality; crosstalk; rectus femoris/hamstrings signals	Demonstrated approaches to identify/mitigate EMG crosstalk, strengthening interpretation of RF activity in SKG.	Observational
Bijleveld-Uitman, M., van de Port, I. G., & Kwakkel, G. (2013)	Adults post-stroke	Rehab/assessment (unspecified)	Controlled comparison	10-m walk test; 6MWT; community walking status	Gait speed and 6MWT were both useful for predicting community ambulation; relative contributions depend on model and context.	Within-subject/Pre–post
Bleyenheuft, C., Deltombe, T., & Detrembleur, C. (2013)	Chronic hemiparetic stroke (n=10)	AFO/orthotic	Controlled comparison	Kinematic segmental covariation (KSC), external work	Dynamic AFO improved paretic-limb kinematic covariation; prefabricated AFO did not.	Within-subject/Pre–post
Bowden, M. G., Balasubramanian, C. K., Neptune, R. R., & Kautz, S. A. (2006)	Chronic hemiparesis (n=47)	Instrumented gait analysis	Not stated	A–P GRFs; % paretic propulsion; walking speed	Paretic propulsion proportion scaled with hemiparetic severity; A–P GRF metrics captured paretic leg contribution.	Observational

Bowden, M. G., Behrman, A. L., Neptune, R. R., Gregory, C. M., & Kautz, S. A. (2013)	Chronic stroke	Rehab/assessment (unspecified)	Not stated	Walking speed, endurance, propulsion symmetry	Responders improved speed (~0.27 m/s) and propulsion symmetry; nonresponders showed limited gains.	Observational
Bregman, D. J. J., Harlaar, J., Meskers, C. G. M., & de Groot, V. (2012)	Neurologic impairment (MS and stroke; n=10)	AFO/orthotic	Controlled comparison	Energy cost, joint kinematics/kinetics, ankle work	Energy-storing AFO reduced walking energy cost (~10%) by partially taking over ankle positive work.	Within-subject/Pre-post
Burpee, J. L., & Lewek, M. D. (2015)	Chronic stroke	Instrumented gait analysis	Not stated	Kinematics/kinetics during failed swing-phase clearance	Characterized movement patterns underlying failed toe clearance, informing mechanisms of tripping risk post-stroke.	Observational
Buurke, J. H., Nene, A. V., Kwakkel, G., Erren-Wolters, C. V., & Hermens, H. J. (2008)	Adults post-stroke (follow-up)	Instrumented gait analysis	Not stated	Joint kinematics/kinetics; spatiotemporal parameters	Showed time-dependent improvements in gait parameters with persistent deficits in push-off and knee flexion.	Observational
Buurke, T. J., Lamoth, C. J. C., van der Woude, L. H. V.,	Adults post-stroke	Instrumented gait analysis	Not stated	Muscle coordination/co-activation; spatiotemporal	Changes in walking speed modulated muscle coordination patterns post-stroke; coordination constraints	Observational

& den Otter, A. R. (2016)				parameters; speed	contributed to limited speed increases.	
Campanini, I., Merlo, A., & Damiano, B. (2013)	Stroke with stiff-knee gait (n=52) + healthy (n=20)	EMG-based assessment	Not stated	PMVA (proxy for push-off), KfV, RF activity	85% of SKG cases driven by inadequate push-off vs RF braking; supports using GA to tailor treatment.	Observational
Chantraine, F., Nguissan, L., Deltombe, T., & Lejeune, T. M. (2005)	Adults with hemiparetic stiff-knee gait	Rehab/assessment (unspecified)	Pre-post / within-subject	Knee flexion in swing; stiff-knee gait parameters; walking speed	Rectus femoris motor branch block improved knee flexion during swing and reduced stiff-knee characteristics in adults with hemiparesis.	Interventional (nerve motor branch block)
Cheung, V. C. K., Turolla, A., Agostini, M., Silvoni, S., Bennis, C., Kasi, P., Paganoni, S., Bonato, P., & Bizzi, E. (2012)	Adults post-stroke	EMG-based assessment	Not stated	Synergy structure; merging; cortical markers	Altered/merged synergies reflect cortical damage and relate to motor performance.	Observational
Clark, D. J., Ting, L. H., Zajac, F. E., Neptune, R. R., & Kautz, S. A. (2010)	Adults post-stroke and controls	EMG-based assessment	Not stated	Synergy number/merging ; walking speed; coordination complexity	Merged synergies post-stroke correlate with reduced performance and coordination complexity.	Observational

Den Otter, A. R., Geurts, A. C. H., Mulder, T., & Duysens, J. (2006)	Post-stroke hemiparesis	Treadmill/velocity training	Pre-post / within-subject	Temporal EMG patterning; coactivity indices	Functional gait recovery occurred without clear normalization of temporal EMG patterns.	Within- subject/Pre- post
Duclos, C., Nadeau, S., Bourgeois, N., Bouyer, L., & Richards, C. L. (2014)	Chronic hemiparesis post-stroke (n=10)	Rehab/assessment (unspecified)	Controlled comparison	Gait speed; cadence; step lengths; joint moments and powers	Ankle loading modestly increased gait speed and paretic step length; increased hip/knee moments and powers without altering stance/swing timing.	Within- subject/Pre- post
Enderby, P., Pandyan, A., Bowen, A., Hearnden, D., Conroy, P., & Langhorne, P. (2017)	Adults post- stroke	Rehab/assessment (unspecified)	Not stated	Service access; workforce; geography; costs	Identifies workforce shortages, cost, and uneven service distribution as key obstacles to post-stroke rehabilitation access.	Systematic review
Harper, N. G., Esposito, E. R., Wilken, J. M., & Neptune, R. R. (2014).	Adults with gait impairment (clinical population)	AFO/orthotic	Not stated	Spatiotemporal parameters; ankle/knee kinematics; energy cost	AFO stiffness meaningfully altered ankle and knee motion and walking economy, underscoring the need to tune AFO properties.	Observational; Within- subject/Pre- post
Everaert, D. G., Thompson, A. K., Chong, S. L., & Stein, R. B. (2010)	Adults with foot drop (incl. stroke)	FES	Not stated	Corticospinal excitability; EMG; gait metrics	Findings suggest FES use can modulate corticospinal connectivity alongside functional improvements.	Intervention (FES; neurophysiology)

Franceschini, M., La Porta, F., Agosti, M., & Massucci, M. (2018)	Adults 1 year post-stroke	Rehab/assessment (unspecified)	Not stated	HRQoL vs neurological impairment indices	Neurological impairment at one year correlates with HRQoL domains, underscoring multi-domain influences beyond motor deficits.	Observational
Fulk, G. D., He, Y., Boyne, P., & Dunning, K. (2017)	Adults post-stroke (n=441)	Rehab/assessment (unspecified)	Controlled comparison	6MWT, Berg Balance Scale, Fugl-Meyer, gait speed	6MWT outperformed gait speed for discriminating home vs community and limited vs full community ambulators; thresholds 0.49 and 0.93 m/s reported for gait speed.	Within-subject/Pre-post
Hsiao, H.-Y., Knarr, B. A., Higginson, J. S., & Binder-Macleod, S. A. (2015a)	Adults with chronic stroke	Speed modulation during treadmill walking	Different walking speeds	Propulsive force; trailing limb angle; ankle moment	Increases in propulsive force were driven predominantly by increases in trailing limb angle rather than ankle moment.	Experimental speed-modulation study
Hsiao, H.-Y., Awad, L. N., Palmer, J. A., Higginson, J. S., & Binder-Macleod, S. A. (2015b)	Adults with chronic stroke (N = 38)	Gait rehabilitation/training analysis	Pre-post / within-subject	Peak propulsive forces (paretic vs nonparetic); walking speed	Changes in paretic propulsion were strongly associated with gains in walking speed following rehabilitation.	Observational longitudinal / within-subject study

Ibala, E. D. (2020)	Adults post-stroke and healthy adults	Self-paced treadmill walking assessment	Fixed-speed treadmill and overground walking	EMG variability; gait adaptability; muscle activation variability	Self-paced treadmill walking demonstrated muscle activity variability patterns more similar to overground walking than fixed-speed treadmill walking, supporting its potential utility for gait adaptability training after stroke.	Doctoral thesis / observational study
Ivanenko, Y. P., d'Avella, A., Poppele, R. E., & Lacquaniti, F. (2004)	Healthy and neurologic populations	Rehab/assessment (unspecified)	Not stated	Muscle synergies; modular control; locomotor adaptation	Describes modular muscle synergy framework underlying locomotion; applied as context for stroke-related motor control changes.	Thesis
Jonkers, I., Delp, S., & Patten, C. (2009)	Lower-functioning adults post-stroke (n = 22) and healthy controls (n = 11)	Instrumented gait analysis during speed modulation	Healthy controls and higher-functioning gait performance	Joint kinematics; hip and ankle power generation; walking speed	Lower-functioning stroke participants demonstrated impaired ability to increase hip and ankle power generation at faster walking speeds, limiting speed modulation capacity.	Observational cross-sectional study
Kerr, A., Durward, B., Lawson, G., & Ryan, S. (2020)	Adults after ischaemic stroke	Rehab/assessment (unspecified)	Randomised comparison	Tibia-to-vertical angle; trailing limb angle;	Tibia-to-vertical angle and related biomechanical variables predicted speed recovery;	Randomised; Within-

	undergoing gait rehabilitation			propulsion; hip extension; walking speed	features can classify responders to gait rehab.	subject/Pre-post
Kerr, A., Rafferty, D., Kerr, K., & Durward, B. (2020)	Community-dwelling stroke survivors	Instrumented gait analysis	Not stated	Tibia-to-vertical angle; walking speed; functional outcomes	Tibia-to-vertical angle significantly correlated with functional walking ability and speed post-stroke.	Observational
Kerrigan, D. C., Gronley, J. K., & Perry, J. (1991)	Adults with spastic stiff-legged gait due to UMN lesions (incl. stroke); n=23	EMG-based assessment	Not stated	Quadriceps/ham strings EMG timing; gait velocity; stride length; heel-rise timing; knee flexion	Inappropriate quadriceps activity during preswing/initial swing common; preswing hamstring activity associated with reduced gait speed and stride length; delayed heel rise prevalent, suggesting plantarflexor weakness.	Observational
Kerrigan, D. C., Roth, R. S., & Riley, P. O. (1998a)	Adults post-stroke with stiff-legged gait (n=5)	Model-based instrumented gait analysis	Simulated torque conditions	Segment torques (hip/knee), simulated knee flexion response	Simulation demonstrated that altering hip torque produced greater changes in knee flexion than altering knee torque, highlighting multi-joint contributors to stiff-knee gait.	Model-based analysis using clinical data

Kerrigan, D. C., Todd, M. K., Della Croce, U., Lipsitz, L. A., & Collins, J. J. (1998b)	Healthy older adults vs younger	Instrumented gait analysis	Younger controls	Knee/hip ROM, gait kinematics	Older adults demonstrated stiff-knee gait characteristics independent of walking speed, suggesting age-related biomechanical constraints relevant to pathological gait.	Observational cross-sectional study
Kerrigan, D. C., Karvosky, M. E., & Riley, P. O. (2001)	Adults post-stroke with spastic paretic stiff-legged gait (n = 20) and able-bodied controls (n = 20)	Instrumented gait analysis	Healthy controls	Joint torques and powers at the hip, knee, and ankle	Stroke participants demonstrated reduced knee power absorption and ankle power generation, alongside abnormal hip, knee, and ankle torques during stance. Findings supported multiple biomechanical mechanisms underlying reduced knee flexion during swing in stiff-knee gait.	Observational cross-sectional study
Knarr, B. A., Kesar, T. M., Reisman, D. S., Binder-Macleod, S. A., & Higginson, J. S. (2013)	Chronic stroke survivors undergoing 12-week FastFES gait retraining	FastFES gait retraining	Pre-post / within-subject	Plantarflexor activation; propulsion; COM acceleration; walking speed	Training improved plantarflexor activation timing and contribution to forward COM acceleration, alongside increases in walking speed and paretic propulsive function.	Within-subject / intervention study

Kollen, B., Kwakkel, G., & Lindeman, E. (2006)	Stroke survivors	Rehab/assessment (unspecified)	Not stated	Clinical/neuro predictors of recovery	Summarized consistent predictors of gait and functional recovery (initial severity, age, etc.).	Systematic review
Kwok, C. S., Potter, J. F., Dalton, G., & Myint, P. K. (2012)	Adults with acute ischaemic stroke	Rehab/assessment (unspecified)	Not stated	Modified Rankin pre-stroke; mortality; length of stay	Pre-stroke disability robustly predicts mortality and LOS, emphasizing baseline status in prognostic models.	Observational
Le, T., & Yang, F. (2015)	Adults post-stroke and/or controls	Treadmill/velocity training	Not stated	Tibialis anterior EMG; foot-clearance metrics	Uneven terrain and speed modulated TA activation and minimum toe clearance, informing tripping risk and training design.	Experimental (terrain × speed)
Lewek, M. D., Raiti, C., & Doty, A. (2018)	Chronic stroke (n=10)	Rehab/assessment (unspecified)	Not stated	Peak/impulse propulsion; TLA; ankle moment	Demonstrated large 'propulsion reserve'; increases linked to TLA rather than PF moment.	Within-subject/Pre-post
Lewek, M. D. (2019)	Adults post-stroke	Rehab/assessment (unspecified)	Not stated	Trailing limb angle; anterior GRF; walking speed	TLA strongly reflected propulsive force generation post-stroke, supporting its use as a clinical gait target metric.	Observational
Mahendran, N., Kuys, S. S., & Brauer, S. G. (2016)	Adults post-stroke (n=42)	Rehab/assessment (unspecified)	Not stated	ActivPAL steps, bout	Walking volume and time increased from 1→3→6 months but remained well below levels	Observational

				frequency/intensity	typically associated with health benefits; most activity occurred in short bouts.	
Mahmood, M., Raza, H., & Dehghani-Sanij, A. A. (2022)	Adults post-stroke	AFO/orthotic	Controlled comparison	Spatiotemporal stability metrics during transitions	Adjustable AFO use improved measures of stability during transitional gait tasks compared with no AFO.	Within-subject/Pre-post
Mazzoli, D., Giannotti, E., Manca, M., Longhi, M., Prati, P., Cosma, M., Ferraresi, G., Morelli, M., Zerbinati, P., Masiero, S., & Merlo, A. (2018)	Post-stroke SKG	EMG-based assessment	Not stated	VI EMG; knee kinematics	Described VI activation patterns in SKG; informs targeting beyond RF.	Observational
Merlo, A., & Campanini, I. (2019)	Chronic stroke with SKG (n=160)	EMG-based assessment	Not stated	PIT_KIN, PIT_EMG, PIT_GA; cost outcomes	GA could reduce inappropriate RF BoNT treatments by up to 81%, with substantial cost savings.	Observational
Mohammadi, R., & Phadke, C. P. (2017)	Chronic stroke (n=19) and healthy controls (n=15)	Treadmill/velocity training	Not stated	Peroneus longus & medial gastrocnemius	Paretic PL activation increased at faster speeds; incline influenced nonparetic PL; timing changes differed by group.	Observational

				sEMG timing/amplitude		
Mulroy, S. J., Eberly, V. J., Gronley, J. K., Weiss, W., & Newsam, C. J. (2010)	Adults post-stroke (n=30)	AFO/orthotic	Not stated	3D kinematics, kinetics, surface EMG	All AFOs increased swing/early stance dorsiflexion. Designs allowing plantar- or dorsiflexion mobility differentially affected knee motion and soleus/tibialis anterior EMG.	Within-subject/Pre-post
Neptune, R. R., Clark, D. J., & Kautz, S. A. (2009)	Healthy and post-stroke	Rehab/assessment (unspecified)	Not stated	Synergy contributions to propulsion/support	Describes how propulsion and body support arise from modular muscle activity; provides a lens for post-stroke deficits.	Thesis
Owen, E. (2004)	Orthotic practice (applied to stroke)	AFO/orthotic	Not stated	SVA; knee extension moments; push-off mechanics	Introduces SVA tuning to optimise knee extension, stability, and push-off; foundational for AFO alignment practice.	Thesis
Owen, E. (2005)	Orthotic practice	Rehab/assessment (unspecified)	Not stated	SVA; knee extension; ankle rocker	Details application of SVA to achieve desired knee moments and efficient ankle rocker progression.	Professional/technical guidance

Owen, E. (2010)	Orthotic practice	AFO/orthotic	Not stated	AFO-FC; rocker profiles; SVA tuning	Describes combining AFO tuning with footwear rocker design to optimise shank progression and push-off.	Technical/Methods
Patterson, K. K., Gage, W. H., Brooks, D., Black, S. E., & McIlroy, W. E. (2008)	Adults post-stroke	Rehab/assessment (unspecified)	Not stated	Temporal/spatial symmetry indices; reliability	Compared symmetry measures; recommended standardization for post-stroke assessment.	Methodological/clinical
Perry, J., Garrett, M., Gronley, J. K., & Mulroy, S. J. (1995)	Adults post-stroke	Rehab/assessment (unspecified)	Not stated	Functional walking categories; gait speed; community ambulation	Introduced a practical classification of walking handicap based on functional walking performance; thresholds support clinical decision-making.	Observational
Peterson, C. L., Cheng, J., Kautz, S. A., & Neptune, R. R. (2010)	Chronic stroke and controls	Instrumented gait analysis	Not stated	Leg extension angle; propulsive impulse	Greater paretic leg extension predicted higher paretic propulsive impulse; hip flexor impulse negatively related.	Observational
Pomeroy, V. M., Rowe, P., Clark, A., Walker, A., Kerr, A., Chandler, E., &	Adults early after stroke	AFO/orthotic; Early casting	Controlled comparison	Walking speed; independence; spatiotemporal gait	Evaluated whether early AFO casting enhances walking recovery; key functional outcomes reported.	Randomised

SWIFT Cast Investigators. (2016)						
Reisman, D. S., Binder-Macleod, S. A., & Higginson, J. S. (2009)	Adults post-stroke	Rehab/assessment (unspecified)	Not stated	Walking speed; propulsion symmetry; gait mechanics	Training at faster velocities improved walking speed and propulsion symmetry, supporting task-specific velocity training.	Intervention study
Roelker, S. A., Bowden, M. G., Kautz, S. A., & Neptune, R. R. (2019)	Adults post-stroke (biomechanics literature)	Rehab/assessment (unspecified)	Not stated	Paretic propulsion; leg extension; plantarflexor module; step length asymmetry	Advocates targeting paretic propulsion as a primary rehabilitation goal; propulsion relates closely to functional walking performance.	Narrative review
Selles, R. W., Li, X., Lin, F., Chung, S. G., Roth, E. J., & Zhang, L.-Q. (2005)	Adults post-stroke with ankle spasticity/contracture	Ankle stretching/device	Not stated	Passive/active ROM, stiffness, reflex excitability, walking speed	Improved ankle ROM and stiffness; comfortable walking speed increased after 4-week program.	Noncontrolled intervention (clinical trial)
Selves, C., Stoquart, G., & Lejeune, T. (2020)	Adults post-stroke (varied)	Rehabilitation/gait recovery review	Narrative review / thesis	Predictors of gait recovery; intervention timing; walking outcomes;	Synthesizes evidence supporting individualized, multimodal rehabilitation matched to recovery stage and functional status. Highlights	Thesis / narrative review

				TWIST algorithm; 6MWT	trunk control and lower-limb motor control as key early predictors of gait recovery and discusses the role of outcome measures such as gait speed and 6MWT in predicting community ambulation.	
Stanhope, V. A., Knarr, B. A., Reisman, D. S., & Higginson, J. S. (2014)	Adults post-stroke (n=21)	Instrumented gait analysis	Not stated	Pelvic hiking; hip abduction (circumduction); knee flexion; ankle dorsiflexion	Faster walkers showed greater knee flexion and ankle dorsiflexion; some achieved speed via large circumduction, indicating multiple compensation profiles.	Observational
Stinear, C. M., Lang, C. E., Zeiler, S., & Byblow, W. D. (2019)	Adults post-stroke (prognostic tools)	Prognostic assessment and rehabilitation prediction tools	Narrative review	PREP2; TWIST; neurophysiology ; trunk control; motor prediction	Reviews validated algorithms and clinical predictors of post-stroke motor and walking recovery, emphasizing the importance of early trunk control, lower-limb strength, neurophysiology, and multimodal assessment in predicting functional outcomes.	Narrative review

Teodoro, J., Fernandes, S., Castro, C., & Fernandes, J. B. (2024)	Adults post-stroke (RCTs, 2013–2023)	Rehab/assessment (unspecified)	Not stated	Gait rehab categories; technology-supported training; conventional vs innovative methods	Identified 13 categories of gait interventions; growing emphasis on technology-assisted approaches alongside conventional therapy.	Observational
Ting, L. H., & McKay, J. L. (2007)	Healthy and neurologic populations	Rehab/assessment (unspecified)	Not stated	Muscle synergy theory; locomotor control	Synthesizes how synergies coordinate muscle groups during posture and locomotion; framework applied to post-stroke gait.	Narrative review
Tyrell, C. M., Roemmich, R. T., et al. (2011)	Adults post-stroke	Treadmill/velocity training	Not stated	Trailing limb angle; anterior ground-reaction force; propulsion	Larger trailing limb angle related to greater paretic propulsion, supporting it as a therapeutic target.	Within-subject/Pre-post
Veerbeek, J. M., Kwakkel, G., van Wegen, E. E. H., Ket, J. C. F., & Heymans, M. W. (2011)	Adults ≤2 weeks post-stroke; ADL outcomes ≥3 months	Rehab/assessment (unspecified)	Not stated	Predictors of ADL; neurological status; upper limb paresis; age	High-quality evidence supports baseline neurological status, upper-limb paresis, and age as predictors of final ADL; many prognostic studies had methodological limitations.	Systematic review; Thesis

Veerbeek, J. M., Van Wegen, E., Harmeling-van der Wel, B., & Kwakkel, G. (2011)	Nonambulatory adults within 72 hours post-stroke (n≈150)	Prognostic clinical assessment	Prospective prognostic cohort study	NIHSS; Motricity Index; Fugl-Meyer; gait outcome at follow-up	Simple early clinical measures predicted later walking ability with good accuracy in initially nonambulatory stroke patients.	Prospective prognostic study
Wagle, J., Farner, L., Flekkøy, K., Wyller, T., Sandvik, L., & Fure, B. (2011)	Adults in stroke rehab	Rehab/assessment (unspecified)	Not stated	Neurocognitive tests; functional outcome scales	Early cognitive status is a strong predictor of 13-month functional outcome after stroke.	Observational; Within-subject/Pre-post
Wang, W., Li, K., Yue, S., Yin, C., & Wei, N. (2017)	Adults post-stroke (n=10) and matched controls (n=10)	EMG-based assessment	Not stated	sEMG RMS & MNF of RF, BF, GS; knee flexion angle range	Reduced knee flexion related to altered EMG magnitude/frequency; GS and BF metrics explained substantial variance in paretic knee flexion.	Observational
Wang, X., Guo, Z., & Yan, T. (2018)	Adults post-stroke	Robotic-assisted	Not stated	RAGT effectiveness; participation; neurophysiology	Summarizes RAGT evidence base and outlines research/implementation gaps.	Narrative review
Wikström, J., Georgoulas, G., Moutsopoulos, T., & Seferiadis, A. (2014)	Adults post-stroke	ML/data mining	Not stated	Classification performance; feature engineering;	Machine-learning approaches can classify impairment levels and detect gait events in stroke; studies were small and methods varied widely.	Systematic review

				EMG/kinematics integration		
Wonsetler, E. C., & Bowden, M. G. (2017)	Adults post-stroke	EMG-based assessment	Not stated	EMG activation; joint moments/power s; spatiotemporal parameters; VO2	Speed gains relate to increases in propulsion (often via trailing limb angle), improved knee flexion/dorsiflexion timing, and aerobic capacity changes.	Systematic review
Woodford, H. J., & Price, C. I. M. (2007)	Adults post-stroke	EMG/kinematic biofeedback	Randomised comparison	Upper/lower limb motor outcomes; walking function; adverse events	Evidence for EMG biofeedback showed mixed but suggestive benefits in specific contexts; heterogeneity and small sample sizes limited certainty.	Randomised; Systematic review
Yang, D. J., Park, S. K., Uhm, Y. H., Park, S. H., Chun, D. W., & Kim, J. H. (2016)	Adults post-stroke (n=55)	EMG-based assessment	Not stated	Quadriceps sEMG (%RVC), limits of stability, gait velocity	Greater quadriceps activation associated with better balance and faster gait; supports strengthening focus in rehab.	Observational
Yuan, H., Ge, P., Du, L., & Xia, Q. (2019)	Adults with hemiplegic stroke (n=30)	Rehab/assessment (unspecified)	Controlled comparison	Rectus femoris–biceps femoris co-contraction ratio; knee extension angle;	Higher antagonist co-contraction on the involved side related to greater knee extension and worse impairment metrics;	Observational

				Fugl-Meyer; MAS	suggests co-contraction role in stance stability.	

Appendix B – Supplementary information for Study 1

B.1 University of Strathclyde Ethics

Ethical Approval Granted for **Paper DEC.BioMed.2022.329**

B.2 Breakdown of Questionnaire Flow

Stroke rehabilitation analysis device

Start of Block: Permission



Q-1 By clicking Accept I consent to my responses being collected and analysed by the University of Strathclyde. I understand that my data will be anonymous and no identifiable data will be collected. I am aware that the responses collected will be used for research in line with ethical approval by the University of Strathclyde. I am aware I cannot remove my responses once the questionnaire has finished but that I can contact the University to discuss overall results once analysed.

Many thanks for your participation.

Rhona Campbell

rhona.campbell.100@strath.ac.uk

☐ Accept (1)

End of Block: Permission

Start of Block: Demographics

Q1 What is your profession?

- ☐ Orthotist (1)
- ☐ Physiotherapist (2)
- ☐ Occupational Therapist (3)
- ☐ Nurse Specialist (4)
- ☐ Doctor (5)
- ☐ Health Care Assistant (HCA) (6)
- ☐ Other (Please explain) (7)

Q2 How many years have you been certified?

- ☐ Less than 1 year (1)
 - ☐ 1 to 4 years (2)
 - ☐ 5 to 9 years (3)
 - ☐ Over 10 years (4)
-

Q3 In what setting do you normally see patients? (Multiple answers accepted)

- ☐ Hospital acute / wards (1)
 - ☐ Hospital outpatient clinics (2)
 - ☐ Community outpatient (3)
 - ☐ Home visits (including care homes) (4)
 - ☐ Other (Please explain) (5)
-

Q4 How long do your sessions with each patient last?

- ☐ Less than 20 minutes (1)
 - ☐ 25 minutes to 1 hour (2)
 - ☐ Over 1 hour (3)
-

Q5 How long would you normally be seeing a patient for treatment or review?

- ☐ Single review sessions only (1)
- ☐ 1 to 3 weeks (2)
- ☐ 4 to 6 weeks (3)
- ☐ Over 6 weeks (4)

End of Block: Demographics

Start of Block: Gait Assessment

Q6 Do you treat patients after a Stroke?

- ☐ No (1)
- ☐ Yes (2)

Skip To: End of Survey If Q6 = No

Q7 Do you normally do a gait assessment as part of your treatment and planning?

- ☐ No (1)
- ☐ Yes (2)

Display This Question:

If Q7 = No

Q8 Why not?

Display This Question:

If Q7 = Yes

Q9 What gait analysis tools do you use? (Multiple answers accepted)

- ☐ Full gait lab (1)
 - ☐ Video analysis (2)
 - ☐ Visual assessment (3)
 - ☐ Other (Please explain) (4)
-

Q10 Which aspects of the gait assessment do you use when planning treatments? (Multiple answers accepted)

- ☐ Gait speed (1)
 - ☐ Step and/or stride length (2)
 - ☐ Muscle activity / EMG (3)
 - ☐ Joint range of motion (4)
 - ☐ Limb velocity (5)
 - ☐ Limb power (6)
 - ☐ Shank inclination (7)
 - ☐ Other (Please explain) (8)
-
- ☐ I don't use gait analysis for this reason (9)

Q11 Do you feel you currently have access to suitable gait analysis for every patient?

☐ No (1)

☐ Yes (2)

Display This Question:

If Q11 = No

Q12 What do you feel are the main limitations to accessing gait analysis for every patient? (Multiple answers accepted)

☐

Time with patient (1)

☐

Lack of skills or training (2)

☐

Lack of equipment (3)

☐

Other (Please explain) (4)

End of Block: Gait Assessment

Start of Block: Gait assessment with AFO

Q13 Do you treat patients who have ankle foot orthoses (AFOs)?

☐ No (1)

☐ Yes (2)

Skip To: Q19 If Q13 = No

Q14 Do you have any input regarding provision or timing of the AFO use with the patient?

- ☐ Yes (1)
- ☐ No (2)
- ☐ Sometimes (please explain) (3)
-

Display This Question:

If Q14 = Yes

Q15 What gait information do you use when prescribing or tuning an AFO? (Multiple answers accepted)

- ☐ Joint range of motion (1)
- ☐ Muscle activity / EMG (2)
- ☐ Gait speed (3)
- ☐ Shank inclination (4)
- ☐ Other (Please explain) (5)
-

Q16 Do you feel there are limitations to gait analysis when wearing an AFO?

- ☐ No (1)
- ☐ Yes (2)

Display This Question:

If Q16 = No

Q17 Please explain why you answered No to Question 23.

Display This Question:

If Q16 = Yes

Q18 Please explain why you answered Yes to Question 23.

Q19 If a device was available to help easily perform gait analysis on patients with or without an AFO do you think this is something you would use more regularly as part of your practice?

- ☐ No (1)
- ☐ Yes (2)
- ☐ Maybe (please explain) (3)

End of Block: Gait assessment with AFO

B.3 Participant Information Sheet (PIS)



Participant Information Sheet

Name of department: Biomedical Engineering

Title of the study: Analysis of current assessment facilities and practices post Stroke

Introduction

My name is Rhona Campbell and I am a PhD student within the department of Biomedical Engineering at the University of Strathclyde, Glasgow, UK. I am undertaking this study as part of my research into Stroke rehabilitation. I am conducting this research under the supervision of the Chief Investigator – Dr Andy Kerr, Senior Lecturer, department of Biomedical Engineering.

What is the purpose of this research?

According data published by the Stroke Association there are more than 100,000 strokes in the UK each year with stroke the fourth largest cause of death in the country. Post-stroke hemiparesis (weakness occurring on one side of the body) is a common consequence of stroke and, for lower limb this weakness generally results in reduced ability to walk. Gait retraining and recovery of 'normal' function is therefore a major focus for rehabilitative treatment post-stroke.

Previous studies within the university have focussed on the development of novel data collection devices which can be worn alongside (underneath) standard orthotic devices such as AFOs. These devices would enable the clinician (or wearer feasibly) to collect EMG and gait kinematic data to help inform treatment decisions or act as rehabilitation outcome measures.

While it is known that gait analysis is an integral part of gait rehabilitation and treatment, access to Gold standard 3D analysis systems such as Vicon is limited. Many clinicians involved in Stroke rehabilitation services rely on visual gait analysis or limited video analysis when making clinical decisions.

This research aims to determine the clinical (marketplace) need for the novel device in development. Whether clinicians themselves feel there is a current need for low cost gait analysis alternatives which would allow better clinical decision making and therapy outcomes for the service users.

Do you have to take part?

No. You do not have to take part as participation is voluntary. If you do not wish to take part in the study, there will be no consequences of doing so. Whether you participate, or not, will have no consequences for your relationship with the University of Strathclyde.

What will you do in the project?

You will begin by following a link to the survey. The consent form is on the first page of the survey. If you give consent to take part in the study you will click the button to continue, and you will be directed on to the survey questions. If you decline to give consent you will be able to navigate out of the survey.

The survey includes questions about your background and training, your experience of with working with people after a Stroke and whether you are also involved in Orthosis provision. The survey will focus on gait analysis and lower limb treatment only. The survey should take no longer than 15 minutes to complete.

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The screen will state "Thank you for completing the survey" to mark the end of the survey and confirm that your responses have been saved.
No monetary incentive will be provided for taking part in the survey.

Who should take part?

We are looking for participants who work with, or have worked with, people affected by Stroke. You should have at least 6 months experience of this work. This may include, but is not limited to, Orthotists, occupational therapists, physiotherapists, nurses, doctors, or surgeons. Participants must be able to read and comprehend English.

What information is being collected in the project?

Information collected includes your experience with gait rehabilitation after a Stroke. Your experience with gait analysis and orthotic provision for the lower limb (such as AFOs, KAFOs and knee braces). None of the information collected will be identifiable.

Who will have access to the information?

The study is conducted in line with the University General Data Protection Regulation (GDPR) Policy. The University Privacy Notice can be found on the following website: <https://www.strath.ac.uk/ethics/information-sheet-and-consent-form/information>. Information that you provide will remain confidential. All data will be stored on a secure platform accessed only by the investigators. Only non-identifiable information will be used when findings from the study are shared.

Where will the information be stored and how long will it be kept for?

Data obtained from this study will be stored on the University's secure platform and accessed only by researchers. As data is anonymous, it may be stored indefinitely.

What happens next?

Thank you for your attention and time in reading the Participant Information.

If you would like to find out more about the project or wish to ask questions before participating, please contact the researchers.

Your responses to the survey will be anonymous.

The results of this study will be reported in the researcher's (Rhona Campbell) PhD thesis and may be published in other academic papers, presentations or conferences. As the survey is anonymous, the identities of participants are unknown to investigators and will therefore not be published.

Researcher contact details:

Rhona Campbell
PhD Student
Department of Biomedical Engineering
University of Strathclyde
Email: Rhona.campbell.100@strath.ac.uk

Chief Investigator details:

Dr Andrew Kerr

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Senior lecturer
Department of Biomedical Engineering
University of Strathclyde
Email: a.kerr@strath.ac.uk

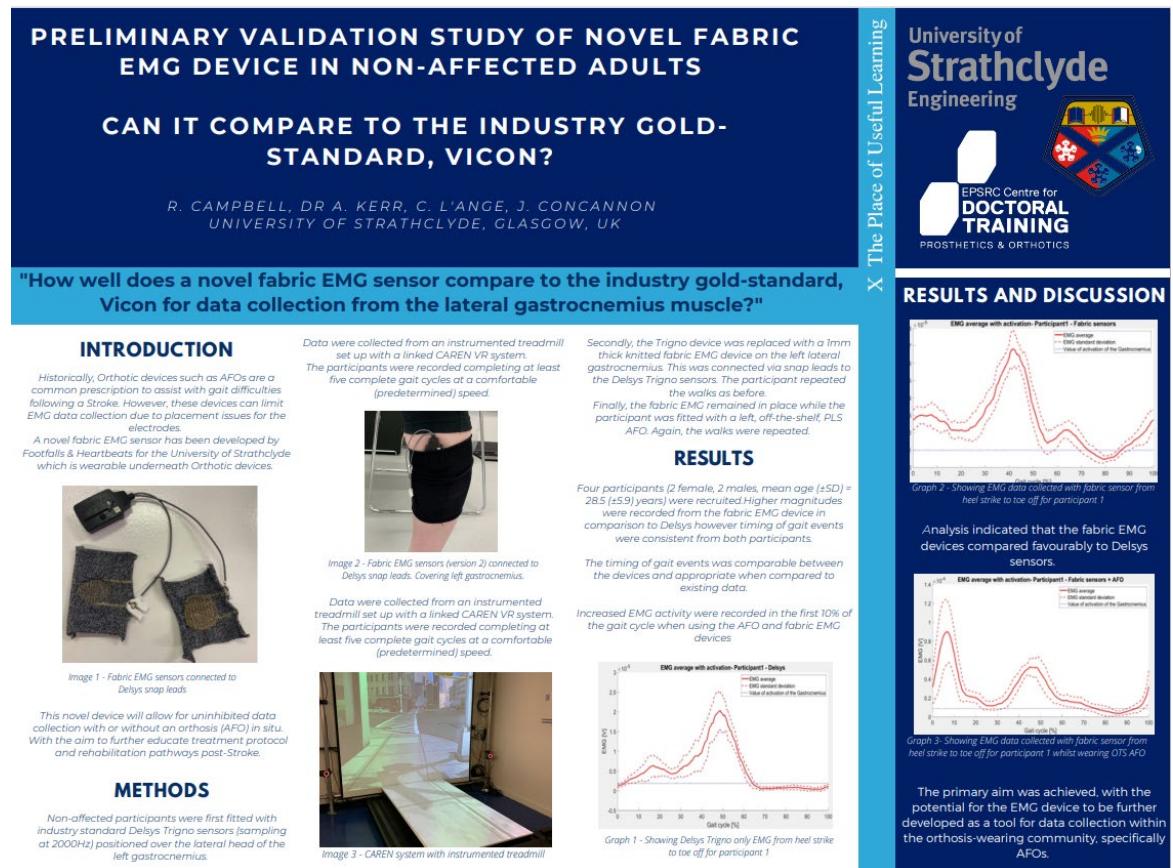
This research was granted ethical approval by the Department of Biomedical Engineering Ethics Committee, University of Strathclyde.

If you have any questions/concerns, during or after the research, or wish to contact an independent person to whom any questions may be directed or further information may be sought from, please contact:

Linda Gilmour
Secretary to the Departmental Ethics Committee
Department of Biomedical Engineering
University of Strathclyde
Tel: 0141 548 3298
Email: linda.gilmour@strath.ac.uk

Appendix C – Supplementary information for Study 2

C.1 Poster Presentation for ESMAC Conference, Dublin, 2022



C.2 University of Strathclyde Ethics

Ethical Approval Granted for **UEC21/54 Kerr/L'Ange/Campbell**

C.3 Participant Information Sheet (PIS)



Participant Information Sheet

Name of department: Department of Biomedical Engineering

Research team: (Chief Investigator) Dr Andrew Kerr, Rhona Campbell (PhD researcher in Prosthetics and Orthotics), Jessica Concanon (Masters researcher Biomedical Engineering).

Title of the study:

"The characteristics and changes of muscle activity and leg movements of able-bodied people during normal gait and speed variation within a lab setting and in a real-world setting."

Introduction

According to the 2018 State of the Nation report by the Stroke Association there are more than 100,000 strokes in the UK each year (NICE, 2019). Post-stroke hemiparesis (weakness occurring on one side of the body) is a common consequence of stroke with up to 85% of the total stroke population affected. For lower limb this weakness generally results in reduced ability to walk (with or without aids) and therefore 'gait retraining' (rehabilitation of walking) is a major focus for treatment post-stroke.

Understanding the differences in the walking pattern between a non-affected person and someone who has survived a Stroke is important for the detection of patterns which may be used to develop more targeted rehabilitation treatments post-Stroke.

In this study we are aiming to use activity monitors to collect measurements of different walking patterns in a controlled environment (in the lab) as well as on a short walk outside. This study will be using people whose walking ability has not been affected by Stroke (or any other medical issues). This information will help us to analyse the data collection monitors for reliability as well as creating a suitable collection of comparable results for future research.

What is the purpose of this research?

This study aims to test the reliability of specialised activity sensors in a variety of situations. Specifically their ability to provide data which can be used to identify changes in walking patterns.

Do you have to take part?

It is up to you whether you take part in the investigation or not. You can refuse to be a participant in the study, and you can withdraw at any time you wish without having to give a reason.

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If you wish to withdraw from the study after you have already taken part, all non-anonymous and pseudo-anonymised data will be cancelled at your request before the data are anonymised.

What will you do in the project?

Interested individuals will be invited to attend a 2-hour appointment at the ~~Motek~~ Gait laboratory in the National Centre for Prosthetics and Orthotics (Curran Building) at the University of Strathclyde which will be arranged via email correspondence or by telephone.

Upon arrival:

You will be asked to wear or change into close-fitting attire of a T-shirt and cycling shorts (or similar) and a pair of well-fitting trainer footwear. Please do not wear loose fitting clothing from the waist down as this will hinder the data collection. Due to current COVID ~~restrictions~~ we ask that you bring your own clothing for the study. We request that you attend the appointment on your own to reduce the number of people within the lab as per COVID guidelines however, if you feel that you will require a chaperone, please contact the researcher team to discuss this.

You will be met by one of the research team and signed into the building. You will be required to follow all University COVID guidelines for hand washing, PPE and face coverings (unless medically exempt). We will be happy to answer any questions you have about the project, and we will familiarise you with the location and changing room if required as well as the protocol. When you are happy, we will ask you to sign a consent form.

Preparation for the testing:

When you are happy to proceed, we will take measurements of your weight, height, leg lengths and width of your pelvis, knees, and ankles. This is needed for the computer to make the calculations as you walk.

The researcher will then place 16 small markers over the skin of specific parts of your body (e.g., knee, ankle and heel) as in figure (1) below. Six special sensors will then be attached to the skin overlying your calf and shin muscles using hypoallergenic tape figure (2), these will record the activity of your muscles. This may require a small area (2cm²) to be shaved. 2 small wireless activity monitors will also be placed on the leg, one on the thigh and one on the calf as in fig (3) below.

After these preparations, which will take approximately 15 minutes, we will proceed with the experiment which has two parts:

Testing part one:

In the ~~Motek~~ gait lab you will be carefully fitted into the safety harness and stood upon the treadmill facing the display screens. The screens will be programmed using the CAREN system to display a simple straight path in front of you which will move in time with your walking on the treadmill.

First, you will be asked to walk at your own comfortable speed for three separate data recordings. Next you will be instructed to await an on-screen instruction to increase (e.g., if they were hurrying to a class) or decrease (e.g., walking alongside an older adult or young child) your speed. This will be repeated three separate times. You will be encouraged to take rests between walks if you need to.

A second CAREN program will be used to simulate a standard UK pedestrian road crossing, you will be asked to walk at your normal speed and stop at the near side of the pedestrian crossing to wait until the signal is given that the crossing is clear to cross. This walk will be repeated three times.

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**Testing part two:**

The second part of the study will capture walking outdoors. You will walk with the researchers to the outside of the building, you will still be wearing the EMG sensors and activity trackers however the reflective markers can be removed at this point if necessary. Once outside of the building you may remove your mask if you wish as long as a minimum 2-meter distance is maintained between you and the researchers.

You will walk a predetermined route around the pedestrian area between the Curran building, Sports Centre and Lord Hope Building. This will comprise of downhill walking, uphill walking and steps. This walk will then be repeated in reverse after a rest period.

Departure:

When all the walks have been completed you will be taken back to the National Centre for Prosthetics and Orthotics and all the markers will be removed, you will then be free to change back into your normal clothes.

You are very welcome to give your thoughts about the experiment.

Why have you been invited to take part?

We are inviting between 15 to 20 able-bodied people over the age of 18 years old to participate using the following criteria:

- 18 years old and over
- Would consider themselves fit and well with no physical restrictions.
- Are able to walk more than 20 steps independently without using a frame or walking stick.
- Are not currently on any medication affecting walking ability or comprehension.
- Are independent in activities of daily living.
- Are able to follow simple instructions and communicate in English.
- Are able to provide informed consent.
- Are able to come to the University of Strathclyde during working hours (9-5) Monday to Friday, for a single two-hour appointment.

To prevent injury or harm we are excluding people from taking part for the following reasons:

- Current COVID symptoms or diagnosis.
- Recent close contact with any person suspected of having COVID.
- Known cardiac problems exacerbated by exercise.
- Currently complaining of joint or muscle problems that affect walking ability.
- Known neurological conditions that affect walking ability.
- Uncorrected hearing or vision problems.
- A history of motion sickness.
- A history of vestibular or balance problems.
- An active skin condition which could be irritated with sticky tape.
- Any condition which restricts their ability to participate in mild to moderate activity
- Any person knowing, or believing they could be, pregnant.

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What are the potential risks to you in taking part?

There should be very little risk from participating in our study. Slipping and tripping during the indoor portion of the study will be minimised by the use of a safety harness and by removing any unneeded obstacles and cleaning and fluid spills. Outdoors efforts will be made to minimise risk by removing any items from the intended walking path and encouraging you to walk at a comfortable speed. Some people may experience motion sickness whilst using the indoor treadmill and display screen (Motek) system, should you feel unwell the trial will stop and you will be encouraged to rest or remove yourself from further testing if you wish. You may also feel some joint or muscle discomfort from the walking we will ask you to do, although this is a small risk it is worth bearing in mind. We will keep checking on how you feel so that if you feel tired, sore or unwell, we will stop the experiment immediately. If you feel unwell during the testing you will be able to take a break, if you feel you are unable to continue your data will be deleted from the system.

There is a small risk of skin irritation due to the attachment of the markers to your skin. If irritation occurs during the data collection, we will stop the trial and remove the markers. If this occurs, or if you experience irritation after leaving the University we suggest cleaning the area with cool water and mild soap and leaving the area uncovered. If irritation continues anti-itch creams may be used or pharmacist advice sought regarding antihistamine use. If the irritation continues for longer than 5 days, the skin becomes red/ swollen or the skin is broken please seek medical advice immediately.

Due to the current COVID [situation](#) there is a risk of COVID transmission. The continuation of research is vital for the University and the wider healthcare community. Therefore, this study has been designed to follow the latest University COVID guidelines as well as local Government guidelines and is only able to proceed with strict procedures in place regarding contact, social distancing and cleaning of equipment before and after use. For more [information](#) please visit the dedicated webpages - www.strath.ac.uk/coronavirus/.

If you have any symptoms of COVID in the week before the study, are tested positive for COVID or have come into close contact with a person suspected of having COVID we ask that you inform us and **DO NOT ATTEND** the University campus. If you develop any COVID symptoms or test positive in the days following the study, please contact the researchers and we will follow necessary protocol as directed by the University and Government. For more information regarding the University COVID guidelines please visit the dedicated webpages - www.strath.ac.uk/coronavirus/

What information is being collected in the project?

At the beginning of the study, we will be recording your height and weight as well as the dimensions of your lower body so that the software we use can provide accurate model constructions. The cameras we use will record how your body moves as you begin walking from a standing position and during a change in speed (increase or decrease). The EMG sensors will measure how much your lower limb muscles are working, and the activity monitors will enable us to compare and analyse the data.

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Who will have access to the information?

The anonymised results of this study may be submitted for presentation at scientific conferences and may be submitted for publication in a peer reviewed journal. Furthermore, the results will be included in the research students' PhD and MSc theses. Any research publications or presentations resulting from this work will only discuss group results and will not report on individuals. At no time will any personal or identifiable information be released.

Where will the information be stored and how long will it be kept for?

All experimental data will be stored securely (on OneDrive) with access limited to the named researchers. Its access and destruction will be in accordance with the University of Strathclyde Data Protection Policy and GDPR. Anonymous data and identifiable consent forms will be kept indefinitely.

Thank you for taking the time to read this information – please ask any questions if you are unsure about what is written here.

What happens next?

If you are interested in being a participant in this study please contact rhona.campbell.100@strath.ac.uk or Jessica.concannon.2020@uni.strath.ac.uk if you would like to participate. Please also complete and sign the consent form on the next page (copies will be available on the testing day also).

If you do not wish to participate then please accept our thanks for taking the time to read this information.

Researchers contact details:

Researcher: Rhona Campbell
Status: PHD student
Department: Biomedical Engineering
Contact: Rhona.campbell.100@strath.ac.uk

Researcher: Jessica Concannon
Status: Masters Student
Department: Biomedical Engineering
Contact: Jessica.concannon.2020@uni.strath.ac.uk

Chief Investigator details:

Researcher: Andrew Kerr
Status: Lecturer
Department: Biomedical Engineering
Contact: a.kerr@strath.ac.uk

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Consent Form

Name of department: Department of Biomedical Engineering

Title of the study: The characteristics and changes of muscle activity and leg movements of able-bodied people during normal gait and speed variation within a lab setting and in a real-world setting.

- I confirm that I have read and understood the Participant Information Sheet for the above project and the researcher has answered any queries to my satisfaction.
- I understand that my participation is voluntary and that I am free to withdraw from the project at any time, up to the point of completion, without having to give a reason and without any consequences.
- I understand that I can request the withdrawal from the study of some personal information and that whenever possible researchers will comply with my request. This includes the following personal data:
 - my personal information from transcripts.
- I understand that anonymised data (i.e. data that do not identify me personally) cannot be withdrawn once they have been included in the study.
- I understand that any information recorded in the research will remain confidential and no information that identifies me will be made publicly available.
- I confirm that I have not tested positive for COVID in the past week, I do not have any current COVID symptoms, and I have not had close contact with any person(s) with a positive diagnosis.
- I consent to being a participant in the project. Yes / No

For research where it has been decided that "no fault compensation" cover will be provided, the following wording needs to be included: In agreeing to participate in this research, I am aware that I may be entitled to compensation for accidental bodily injury, including death or disease, arising out of the research without the need to prove fault. However, such compensation is subject to acceptance of the Conditions of Compensation, a copy of which is available on request.

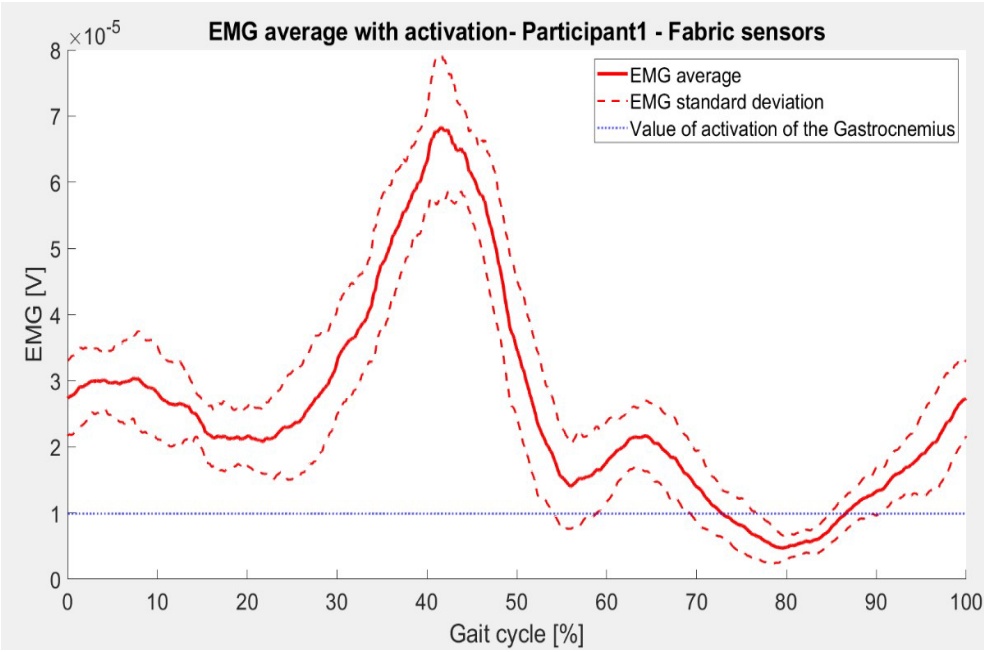
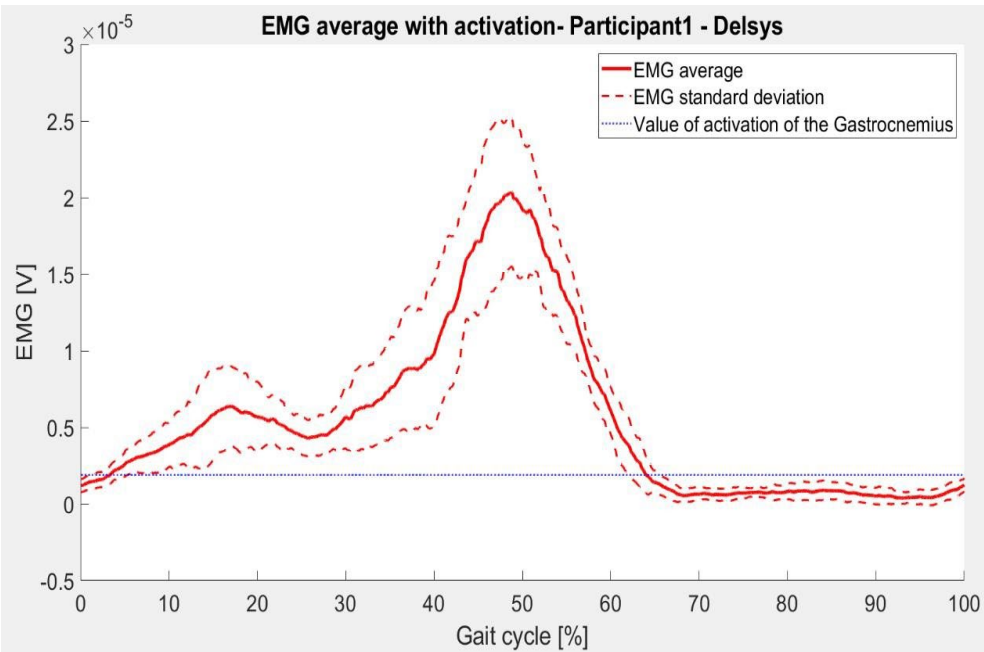
(PRINT NAME)	
Signature of Participant:	Date:

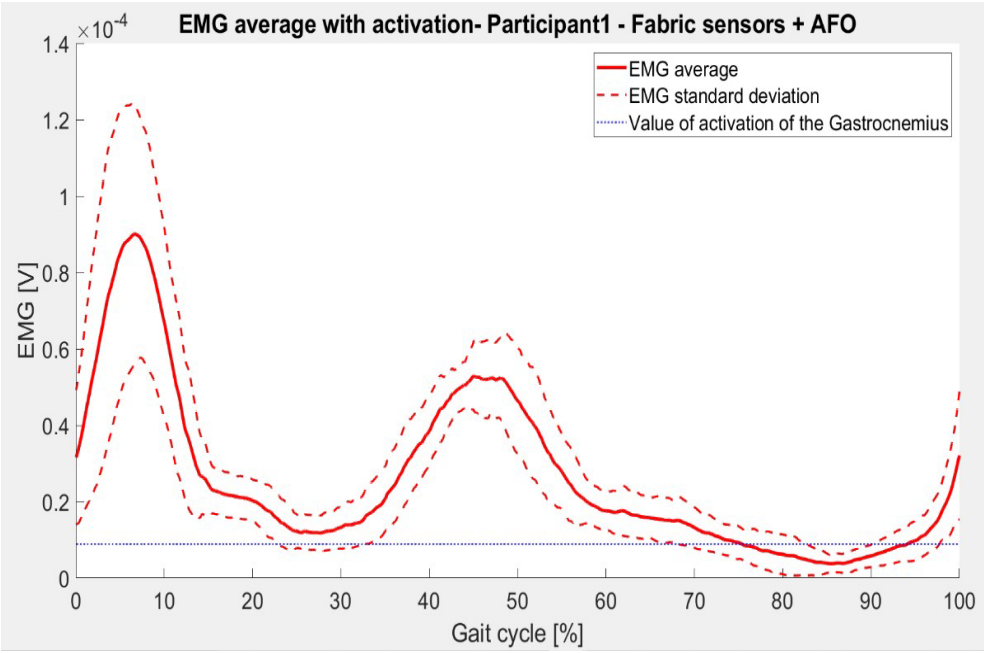
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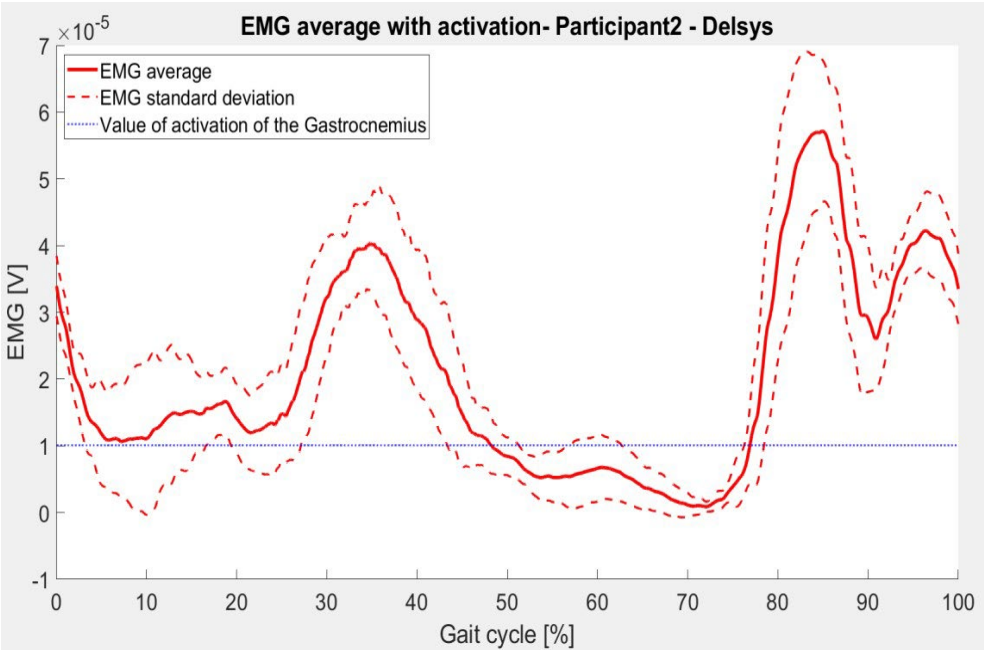
C.4 EMG Graphs – Full Size

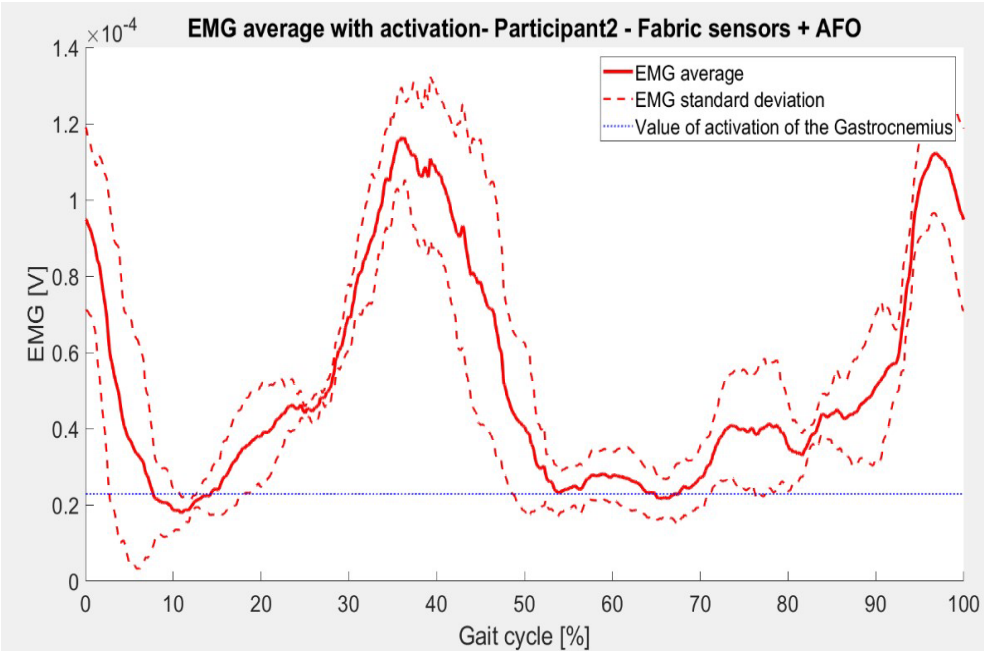
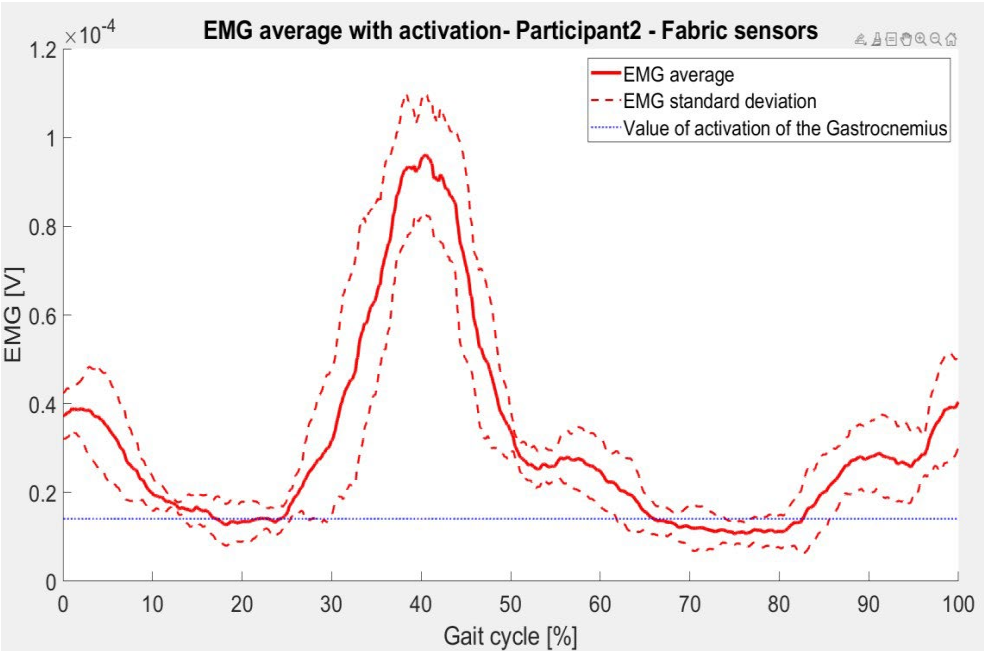
Participant 1



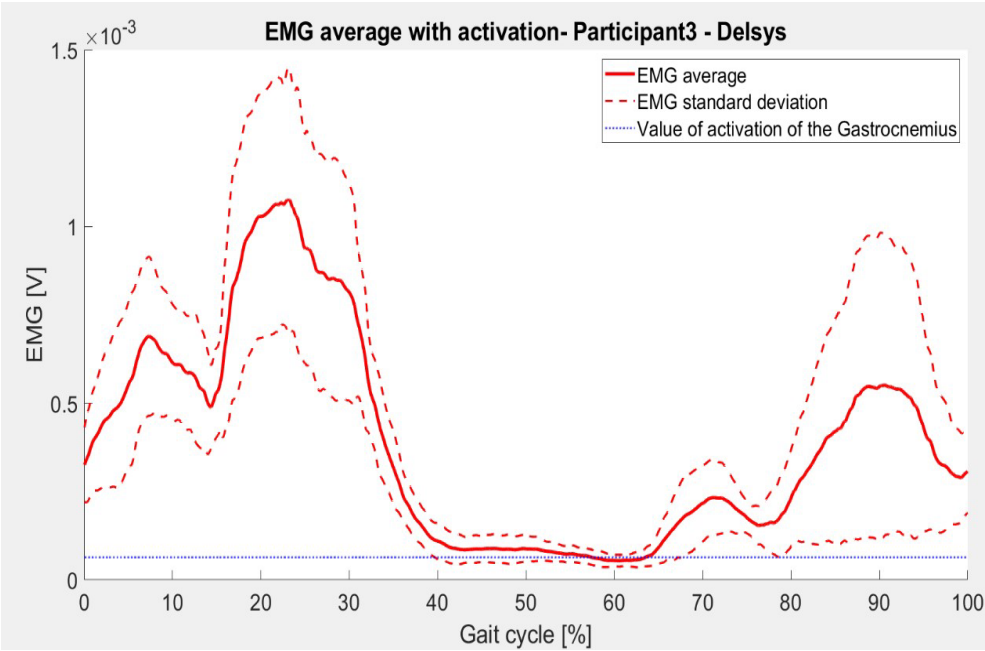


Participant 2

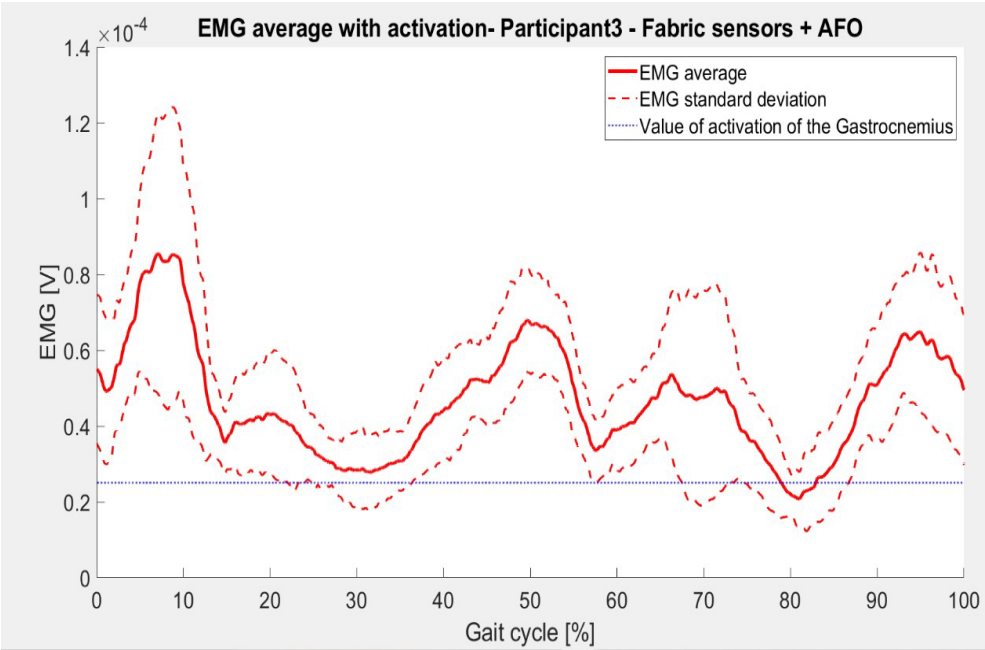




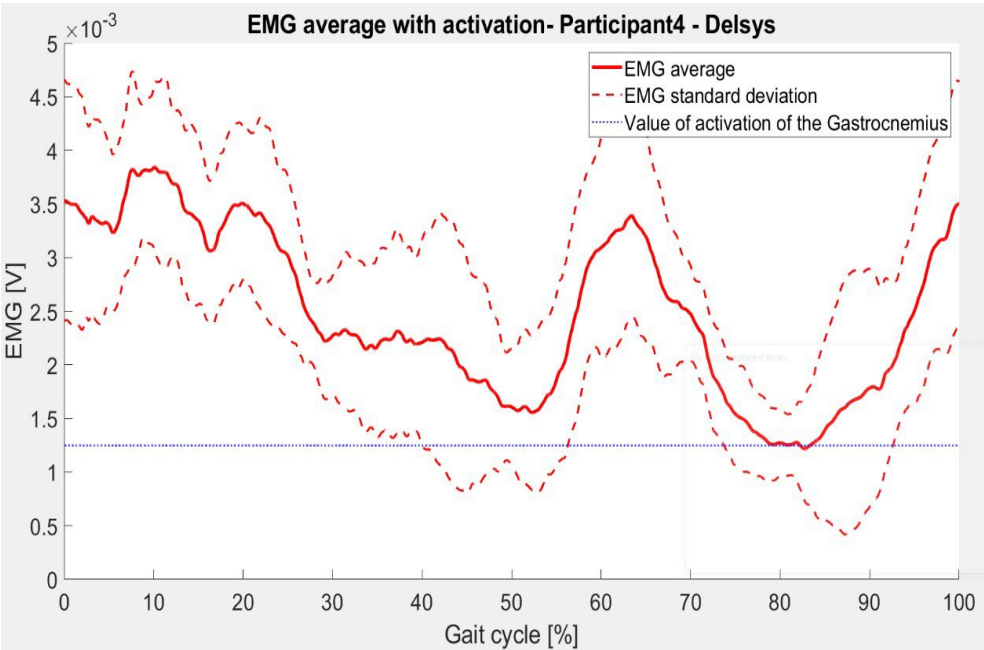
Participant 3



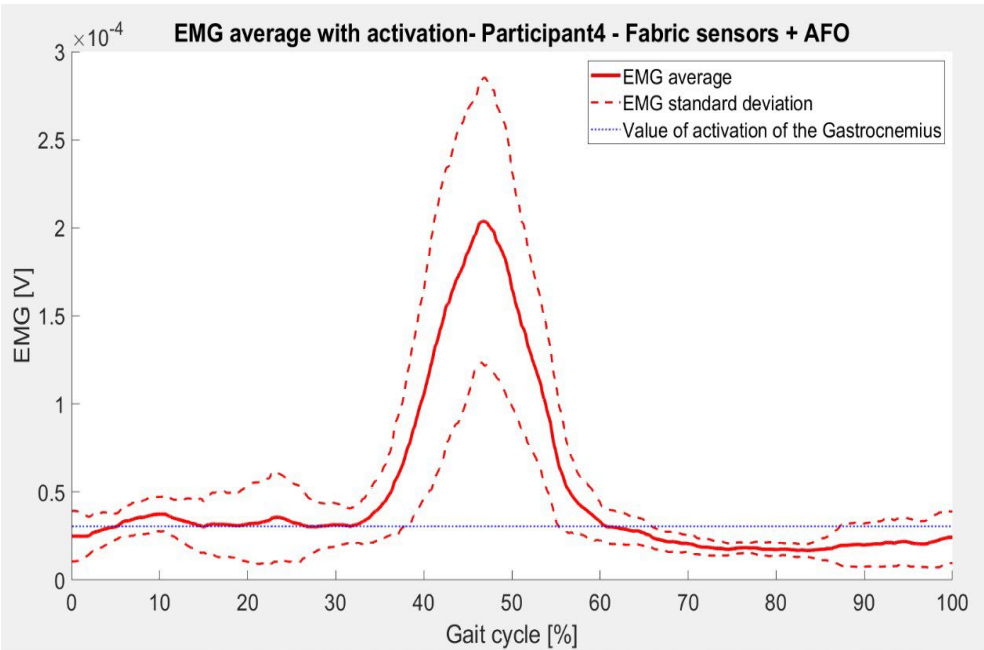
Data for Trial 2 N/A



Participant 4

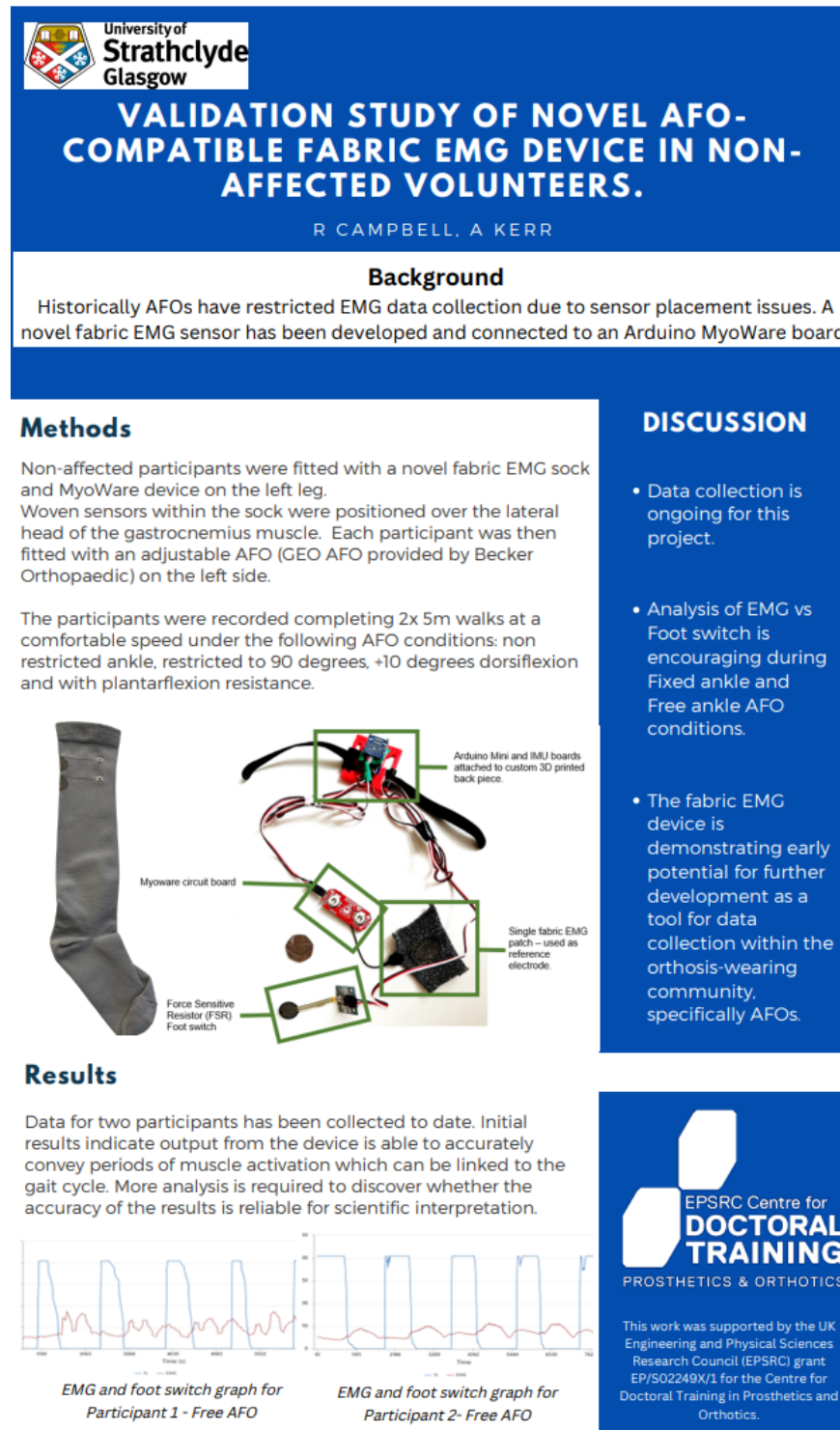


Data for Trial 2 N/A



Appendix D – Supplementary information for Study 3

D.1 Poster Presentation for UK Stroke Forum, Liverpool, 2022



D.2 Participant Information Sheet (PIS)



Participant Information Sheet

Name of department: Department of Biomedical Engineering

Research team: (Chief Investigator) Dr Andrew Kerr, Rhona Campbell (PhD researcher in Prosthetics and Orthotics)

Title of the study:

“Testing of new device to monitor characteristics and changes of muscle activity and leg movements Stroke-affected people with specific ankle-foot movements within a lab setting.”

Introduction

According to the 2018 State of the Nation report by the Stroke Association there are more than 100,000 strokes in the UK each year (NICE, 2019). Post-stroke hemiparesis (weakness occurring on one side of the body) is a common consequence of stroke with up to 65% of the total stroke population affected. For lower limb this weakness generally results in reduced ability to walk (with or without aids) and therefore ‘gait retraining’ (rehabilitation of walking) is a major focus for treatment post-stroke.

Understanding the differences in the walking pattern of someone who has survived a Stroke is important for the detection of patterns which may be used to develop more targeted rehabilitation treatments post-Stroke.

In this study we are aiming to use a novel fabric sock with sewn-in muscles sensors and an activity monitor to collect measurements of different walking patterns in a controlled environment (in the lab) whilst wearing a device that can be altered to restrict or change ankle movement. This study will be involving people whose walking ability has likely been impacted by a previous Stroke event. This information will help us to analyse the data collection monitors for reliability as well as creating a suitable collection of comparable results for future research.

What is the purpose of this research?

This study aims to test the reliability of a novel muscle activity monitor in a variety of situations. Specifically their ability to provide data which can be used to identify changes in walking patterns for people with possibly Stroke-affected walking when wearing an alterable ankle foot device (AFO).

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Why have you been invited to take part?

We are inviting between 5 to 10 people who have previously had a Stroke and who are over the age of 18 years old to participate using the following criteria:

- 18 years old and over
- Have previously been diagnosed with a Stroke affecting left or right side of the body.
- Are able to walk more than 20 steps independently without using a frame or walking stick. Use of a prescribed AFO is allowed if normally worn for walking.
- Are not currently on any medication negatively affecting walking ability or comprehension.
- Consider themselves fairly independent in activities of daily living, assistance of a relative or carer is accepted.
- Are able to follow simple instructions and communicate in English.
- Are able to provide informed consent.
- Are able to come to the University of Strathclyde during working hours (9-5) Monday to Friday, for a single two-hour appointment.
- Sensation intact through limbs enough to register discomfort if present.
-

To prevent injury or harm we are **excluding** people from taking part for the following reasons:

- Current COVID symptoms or diagnosis.
- Recent close contact with any person suspected of having COVID.
- Known cardiac problems exacerbated by exercise.
- Currently experiencing excessive joint or muscle problems that affect walking ability or would be exacerbated through exercise. Pain which is not 'normal' for the participant.
- Known neurological conditions that affect walking ability (outside of Stroke diagnosis).
- Uncorrected hearing or vision problems.
- A history of motion sickness.
- A history of vestibular or balance problems.
- An active skin condition which could be irritated with sticky tape.
- Any condition which restricts their ability to participate in mild to moderate activity
- Any person knowing, or believing they could be, pregnant.
- Any person weighing over 20 stone (127kg)
- Any person with foot size exceeding UK ~~men's~~ 10

Suitability of AFO device for participant will be assessed by main researcher (HCPC registered Orthotist). Significant muscle contractions, spasticity, proximal weakness or alignment issues not otherwise stated will be taken into account and the participant will be advised whether they can continue with the trial.

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Do you have to take part?

It is up to you whether you take part in the investigation or not. You can refuse to be a participant in the study, and you can withdraw at any time you wish without having to give a reason. Participation in the study will not affect your continuation or participation in the Stroke rehab gym sessions. Participation will not affect any of your external rehabilitation or medical treatments.

If you wish to withdraw from the study after you have already taken part, all non-anonymous and pseudo-anonymised data will be cancelled at your request before the data are anonymised.

What will you do in the project?

Interested individuals will be invited to attend a 2-hour appointment at the main [gait](#) laboratory (WC106), level 1 of the Wolfson building at the University of Strathclyde which will be arranged via email correspondence or by telephone. This will likely coincide with your weekly rehab gym sessions if suitable for you.

Please bring with you suitable clothing (described below) and a pair of comfortable sports-trainers. The trainers should standard design not have been altered for any specific purpose. Clothing can be provided by the lab but your own is best.

Upon arrival:

You will be asked to wear or change into close-fitting attire of a T-shirt and cycling shorts (or similar) and a pair of well-fitting trainer footwear. Please do not wear loose fitting clothing from the waist down as this will hinder the data collection. We request that you attend the appointment on your own to reduce the number of people within the lab as per COVID guidelines however, if you feel that you will require a chaperone, please contact the researcher team to discuss this. Anyone who regularly attends your rehab sessions will be welcome to attend the session, this is encouraged if you [feel](#) you would require physical assistance such as when changing or using the toilet.

You will be met by one of the research team and signed into the building. You will be required to follow all University and Scottish Government COVID guidelines for hand washing, PPE and face coverings (unless medically exempt). We will be happy to answer any questions you have about the project, and we will familiarise you with the location and changing room if required as well as the protocol. When you are happy, we will ask you to sign a consent form.

Preparation for the testing:

When you are happy to proceed, we will take measurements of your weight, height, leg lengths and width of your pelvis, knees, and ankles. This is needed for the computer to make the calculations as you walk. During this period, we will also be assessing whether it will be suitable to use the ankle-foot device for you. There are many reasons why it might not be appropriate. If it is decided that the device cannot be used, we will let you know why and the session will be ended. Your range of motion and muscle length at the foot, ankle and knee will be tested (by bending the foot up and down) as well as any instability.

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The researcher will then place 16 small markers over the skin of specific parts of your body (e.g., knee, ankle and heel) as in figure (1) below. A fabric EMG sensor will then be attached to your leg below the knee (this is basically a sock and will not feel any different to a standard thick sock) and the corresponding collection device will be attached (2), this will record the activity of your muscles. This may require a small area (2cm^2) to be shaved. A small activity monitor will also be placed on the leg as in fig (3) below.

Finally, you will be fitted with an Ankle Foot Orthosis (AFO) which has an adaptable hinge (4). This will be worn within your own footwear and should not cause any discomfort.

After these preparations, which will take approximately 30 minutes, we will proceed with the experiment.



Figure 2 - Becker Orthopaedic AFO

Data collection:

In the gait lab you will be shown the area in which you will be asked to walk and given time to become familiar with it.

First, you will be asked to walk at your own comfortable speed with the AFO set to a 'free' ankle meaning there should be minimal change to your normal walk. Each walk will be recorded, and you

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may be asked to repeat the walk if insufficient data was collected.
You will be encouraged to take rests between walks if you need to.

The AFO will next be altered to a fixed ankle position meaning you will be unable to flex or extend your ankle when walking. You will be allowed a short time to become accustomed to this before data collection resumes. You will be asked to repeat your previous walk.

The AFO will then be altered to allow you a small amount of dorsiflexion (lifting the foot upwards), again you will be allowed time to become accustomed to this before repeating the walking.

Finally, the AFO will be altered to allow small amounts of up and downward movement but a spring will be introduced which will actively try to encourage your foot to lift upwards. Again, you will have some time to acclimate to this before repeating the walks.

Departure:

When all the walks have been completed all the markers and the AFO will be removed, you will then be free to change back into your normal clothes, sign out and leave the building.

You are very welcome to give your thoughts about the experiment.

What are the potential risks to you in taking part?

There should be very little risk from participating in our study. Slipping and tripping will be minimised by removing any unneeded obstacles and cleaning any fluid spills. If you feel unstable a second person may accompany you during the walk (without giving direct aid if possible). Persons with known instability may be required to wear a harness connected to the ceiling rail. This will prevent a full fall and any injury should you become unbalanced.

Due to your previous Stroke, you may be more at risk of having a repeat Stroke. Should you feel unwell the trial will stop and you will be encouraged to rest or remove yourself from further testing if you wish. Any acute illness during the trial will be dealt with following University procedure, the researcher is fully First Aid at Work trained and has access to a defibrillator if needed.

You may also feel some joint or muscle discomfort from the walking we will ask you to do, although this is a small risk it is worth bearing in mind. We will keep checking on how you feel so that if you feel tired, sore or unwell, we will stop the experiment immediately. If you feel unwell during the testing you will be able to take a break, if you feel you are unable to continue your data will be deleted from the system.

The ankle-foot device (AFO) may be unfamiliar to you. Care will be taken to ensure it is fitted as comfortably as possible and you will have some time to get used to it before the trial starts. If you find you are too uncomfortable, the device will be removed and you can stop the trial. Additionally, if you

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feel your walking is better with the device, details can be given that you can pass onto your own Orthotist in the future.

There is a small risk of skin irritation due to the attachment of the markers to your skin. If irritation occurs during the data collection, we will stop the trial and remove the markers. If this occurs, or if you experience irritation after leaving the University we suggest cleaning the area with cool water and mild soap and leaving the area uncovered. If irritation continues anti-itch creams may be used or pharmacist advice sought regarding antihistamine use. If the irritation continues for longer than 5 days, the skin becomes red/ swollen or the skin is broken please seek medical advice immediately.

Due to the current COVID [situation](#) there is a risk of COVID transmission. The continuation of research is vital for the University and the wider healthcare community. Therefore, this study has been designed to follow the latest University COVID guidelines as well as the Scottish Government guidelines and is only able to proceed with strict procedures in place regarding contact, social distancing and cleaning of equipment before and after use. For more [information](#) please visit the dedicated webpages - www.strath.ac.uk/coronavirus/.

If you have any symptoms of COVID in the week before the study, are tested positive for COVID or have come into close contact with a person suspected of having COVID we ask that you inform us and **DO NOT ATTEND** the University campus. If you develop any COVID symptoms or test positive in the days following the study, please contact the researchers and we will follow necessary protocol as directed by the University and Government. For more information regarding the University COVID guidelines please visit the dedicated webpages - www.strath.ac.uk/coronavirus/

What information is being collected in the project?

At the beginning of the study, we will be recording your height and weight as well as the dimensions of your lower body so that the software we use can provide accurate model constructions. The cameras we use will record how your body moves as you begin walking from a standing position and during a change in speed (increase or decrease). These cameras can only collect light signals reflected from the markers and do not collect video data. The EMG sensors will measure how much your lower limb muscles are working, and the activity monitors will enable us to compare and analyse the data.

Who will have access to the information?

The anonymised results of this study may be submitted for presentation at scientific conferences and may be submitted for publication in a peer reviewed journal. Furthermore, the results will be included in the research students' PhD thesis. Any research publications or presentations resulting from this work will only discuss group results and will not report on individuals. At no time will any personal or

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identifiable information be released. Its access and destruction according to the University of Strathclyde Data Protection Policy and GDPR

Where will the information be stored and how long will it be kept for?

All experimental data will be stored securely (on OneDrive) with access limited to the named researchers. Its access and destruction will be in accordance with the University of Strathclyde Data Protection Policy and GDPR. Anonymous data and identifiable consent forms will be kept indefinitely.

Thank you for taking the time to read this information – please ask any questions if you are unsure about what is written here.

Privacy notice

Please use this link to access the Privacy Notice for research conducted by the University of Strathclyde. If you have any issues, please feel free to contact the researcher to discuss these.
[Information Sheet and Consent Form - University of Strathclyde](#).

What happens next?

If you are interested in being a participant in this [study](#) please contact rhona.campbell.100@strath.ac.uk if you would like to participate. Please also complete and sign the consent form on the next page (copies will be available on the testing day also).

If you do not wish to participate then please accept our thanks for taking the time to read this information.

Researchers contact details:

Researcher: Rhona Campbell
Status: PHD student
Department: Biomedical Engineering
Contact: Rhona.campbell.100@strath.ac.uk

Chief Investigator details:

Researcher: Andrew Kerr
Status: Lecturer

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Consent Form

Name of department: Department of Biomedical Engineering

Title of the study: Testing of new device to monitor characteristics and changes of muscle activity and leg movements of Stroke-affected people during normal gait within a lab setting.

- I confirm that I have read and understood the Participant Information Sheet for the above project and the researcher has answered any queries to my satisfaction.
- I understand that my participation is voluntary and that I am free to withdraw from the project at any time, up to the point of completion, without having to give a reason and without any consequences.
- I understand that I can request the withdrawal from the study of some personal information and that whenever possible researchers will comply with my request. This includes the following personal data:
 - my personal information from transcripts.
- I understand that anonymised data (i.e. data that do not identify me personally) cannot be withdrawn once they have been included in the study.
- I understand that any information recorded in the research will remain confidential and no information that identifies me will be made publicly available.
- I confirm that I have not tested positive for COVID in the past week, I do not have any current COVID symptoms, and I have not had close contact with any person(s) with a positive diagnosis.
- I consent to being a participant in the project.

(PRINT NAME)

Signature of Participant:

Date:

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D.3 Gait event detection code for Arduino system

```

# --- INSTALLS (Colab): orientation fusion for SVA (Madgwick) ---
# In Colab, run the following in its own cell:
# !pip -q install ahrs
# --- IMPORTS ---
import numpy as np
import pandas as pd
import matplotlib.pyplot as plt
from scipy.signal import butter, filtfilt
from ahrs.filters import Madgwick
# ===== USER CONFIG =====
# CSV input must contain time (s), FSR (arbitrary units), IMU (ax, ay, az,
# gx, gy, gz), and EMG channels.
CSV_PATH = "/content/session.csv" # <-- set your file
COL = {
    "t": "time", # seconds
    "fsr": "fsr_heel", # heel FSR column name
    "ax": "ax", "ay": "ay", "az": "az", # accelerometer (g)
    "gx": "gx", "gy": "gy", "gz": "gz", # gyro (deg/s or rad/s; see
GYRO_IS_DPS)
    "emg": ["emg_ta"] # list of EMG channels; empty [] if
none
}
FS = 200.0 # signal sampling rate (Hz) for all channels (set to your
rate)
GYRO_IS_DPS = True # True if gyros are in deg/s; if rad/s, set False
# FSR event detection
FSR_LP_CUTOFF = 20.0 # low-pass cutoff (Hz) for FSR smoothing
FSR_AUTOTHRESH_PCT = 0.35 # threshold as fraction of [min,max] per-walk
(0..1)
FSR_MIN_BOUT_MS = 100.0 # minimum stance duration to accept (ms)
FSR_DEBOUNCE_MS = 40.0 # refractory after event detection (ms)
# Lag correction from FSR-detected heel contact to true IC
# Based on loading-response being ~0-10% GC; default 7% GC with clamping to
30..100 ms
LAG_PCT_GC = 0.07
LAG_MIN_MS = 30.0
LAG_MAX_MS = 100.0
# EMG processing
EMG_BP = (20.0, 450.0) # band-pass (Hz)
EMG_ENV_LP = 6.0 # envelope low-pass (Hz)
# SVA smoothing for plots
SVA_LP = 20.0 # match your prior 20 Hz zero-phase low-pass
# Time-normalisation
N_GC = 101 # samples per gait cycle (0..100% GC)
# ===== HELPERS =====
def butter_filter(sig, fs, cutoff, btype="low", order=4):
    ny = 0.5*fs
    Wn = np.array(cutoff)/ny if hasattr(cutoff, "__len__") else cutoff/ny

```

```

b, a = butter(order, Wn, btype=btype)
return filtfilt(b, a, sig)
def detect_heel_contacts_from_fsr(t, fsr_raw, fs,
                                lp_cut=FSR_LP_CUTOFF,
                                thr_pct=FSR_AUTOTHRESH_PCT,
                                min_bout_ms=FSR_MIN_BOUT_MS,
                                debounce_ms=FSR_DEBOUNCE_MS):
    """Return indices of heel-contact events from a heel FSR using
    hysteresis."""
    x = butter_filter(fsr_raw, fs, lp_cut, "low")
    x_min, x_max = np.percentile(x, 1), np.percentile(x, 99)
    thr = x_min + thr_pct*(x_max - x_min)
    above = x > thr
    # Find rising edges with debounce
    idx = []
    i = 1
    deb = int(debounce_ms/1000.0*fs)
    min_bout = int(min_bout_ms/1000.0*fs)
    while i < len(x):
        if (not above[i-1]) and above[i]:
            # require that we actually stay above threshold for min_bout to
            avoid noise
            j = i
            while j < len(x) and above[j]:
                j += 1
            if j - i >= min_bout:
                idx.append(i)
                i = j + deb
                continue
            i += 1
    return np.array(idx, dtype=int), x, thr
def apply_ic_lag_correction(ic_idx, t, lag_pct_gc=LAG_PCT_GC,
                           lag_min_ms=LAG_MIN_MS, lag_max_ms=LAG_MAX_MS):
    """Shift FSR-detected heel contacts earlier by a stride-adaptive
    lag."""
    if len(ic_idx) < 2:
        return ic_idx.copy(), np.zeros(len(ic_idx))
    ic_corr = []
    lags_ms = []
    for k in range(len(ic_idx)-1):
        i0, i1 = ic_idx[k], ic_idx[k+1]
        stride_T = t[i1] - t[i0] # s
        lag_s = np.clip(lag_pct_gc*stride_T, lag_min_ms/1000.0,
        lag_max_ms/1000.0)
        lag_samples = int(round(lag_s / (t[1]-t[0])))
        ic_corr.append(max(0, i0 - lag_samples))
        lags_ms.append(lag_s*1000.0)
    # For last IC (no stride duration yet), reuse previous lag

```

```

if len(lags_ms) > 0:
    last_lag_s = np.clip(lag_pct_gc*(t[ic_idx[-1]] - t[ic_idx[-2]]),
                        lag_min_ms/1000.0, lag_max_ms/1000.0)
    last_lag_samples = int(round(last_lag_s/(t[1]-t[0])))
    ic_corr.append(max(0, ic_idx[-1] - last_lag_samples))
    lags_ms.append(last_lag_s*1000.0)
return np.array(ic_corr, dtype=int), np.array(lags_ms)
def fuse_orientation_madgwick(ax, ay, az, gx, gy, gz, fs, beta=0.05):
    """Return quaternion array and pitch angle (deg) relative to
    gravity."""
    # Convert gyro to rad/s if provided in deg/s
    if GYRO_IS_DPS:
        gx, gy, gz = np.deg2rad(gx), np.deg2rad(gy), np.deg2rad(gz)
    f = Madgwick(beta=beta, frequency=fs)
    Q = np.zeros((len(ax), 4))
    q = np.array([1.0, 0.0, 0.0, 0.0])
    for i in range(len(ax)):
        q = f.updateIMU(q, gyr=[gx[i], gy[i], gz[i]], acc=[ax[i], ay[i],
        az[i]])
        Q[i] = q
    # Compute shank pitch relative to global vertical:
    # Get gravity in sensor frame from quaternion, then tilt angle of
    sensor's longitudinal axis vs vertical.
    # Assume sensor long axis ~ tibial axis aligned with +z_s (change if
    needed).
    # Direction cosine matrix from quaternion:
    def quat_to_dcm(q):
        w, x, y, z = q
        return np.array([
            [1-2*(y*y+z*z), 2*(x*y - z*w), 2*(x*z + y*w)],
            [2*(x*y + z*w), 1-2*(x*x+z*z), 2*(y*z - x*w)],
            [2*(x*z - y*w), 2*(y*z + x*w), 1-2*(x*x+y*y)]
        ])
    # Sensor z-axis unit vector in world frame for tilt vs vertical z_w:
    sva_deg = np.zeros(len(ax))
    for i in range(len(ax)):
        C = quat_to_dcm(Q[i]) # maps sensor->world
        z_s_world = C[:,2] # sensor Z in world frame
        # Angle between sensor Z and world vertical (0,0,1)
        cosang = np.clip(z_s_world[2], -1.0, 1.0)
        ang = np.degrees(np.arccos(cosang))
        sva_deg[i] = ang
    return Q, sva_deg
def process_emg(emg, fs, bp=EMG_BP, env_lp=EMG_ENV_LP):
    """Band-pass, rectify, envelope."""
    emg_bp = butter_filter(emg, fs, bp, "bandpass", order=4)
    emg_rect = np.abs(emg_bp)

```

```

env = butter_filter(emg_rect, fs, env_lp, "low", order=4)
return env
def segment_and_normalise(t, y, ic_idx_corr, n_gc=N_GC):
    """Return list of cycles (each resampled to n_gc points) and an (n_gc,)
    mean."""
    cycles = []
    for k in range(len(ic_idx_corr)-1):
        i0, i1 = ic_idx_corr[k], ic_idx_corr[k+1]
        seg_t = t[i0:i1+1]
        seg_y = y[i0:i1+1]
        if len(seg_y) < 5: continue
        # normalise to 0..100% via linear interpolation
        g = np.linspace(0, 1, len(seg_y))
        G = np.linspace(0, 1, n_gc)
        cycles.append(np.interp(G, g, seg_y))
    if cycles:
        M = np.vstack(cycles)
        return cycles, np.nanmean(M, axis=0)
    return [], np.full(n_gc, np.nan)
# ===== LOAD DATA =====
df = pd.read_csv(CSV_PATH)
t = df[COL["t"]].to_numpy().astype(float)
fsr = df[COL["fsr"]].to_numpy().astype(float)
ax, ay, az = df[COL["ax"]].to_numpy(), df[COL["ay"]].to_numpy(),
df[COL["az"]].to_numpy()
gx, gy, gz = df[COL["gx"]].to_numpy(), df[COL["gy"]].to_numpy(),
df[COL["gz"]].to_numpy()
emg_envs = {}
for ch in COL["emg"]:
    emg_envs[ch] = process_emg(df[ch].to_numpy().astype(float), FS)
# ===== FSR-BASED EVENTS + LAG CORRECTION =====
ic_idx_raw, fsr_filt, fsr_thr = detect_heel_contacts_from_fsr(t, fsr, FS)
ic_idx_corr, per_stride_lag_ms = apply_ic_lag_correction(ic_idx_raw, t)
print(f"Detected {len(ic_idx_raw)} heel contacts (FSR). After lag-
correction: {len(ic_idx_corr)}.")
if len(per_stride_lag_ms):
    print(f"Median lag applied: {np.median(per_stride_lag_ms):.1f} ms "
          f"(min {np.min(per_stride_lag_ms):.1f}, max
{np.max(per_stride_lag_ms):.1f})")
# ===== SVA FROM IMU =====
_, sva_raw = fuse_orientation_madgwick(ax, ay, az, gx, gy, gz, FS,
beta=0.05)
sva = butter_filter(sva_raw, FS, SVA_LP, "low", order=4)
# ===== SEGMENT & NORMALISE =====
sva_cycles, sva_mean = segment_and_normalise(t, sva, ic_idx_corr, N_GC)
emg_means = {}
for ch, env in emg_envs.items():
    _, emg_mean = segment_and_normalise(t, env, ic_idx_corr, N_GC)

```

```

    emg_means[ch] = emg_mean
# ===== PLOTS =====
# FSR + events
plt.figure(figsize=(10,3))
plt.plot(t, fsr_filt, label="FSR (LP)")
plt.axhline(fsr_thr, ls="--", lw=1, label="thr")
plt.scatter(t[ic_idx_raw], fsr_filt[ic_idx_raw], s=18, label="IC raw")
plt.scatter(t[ic_idx_corr], fsr_filt[ic_idx_corr], s=18, label="IC
corrected")
plt.xlabel("Time (s)"); plt.ylabel("FSR (a.u.)"); plt.legend();
plt.title("Heel FSR & heel-contact events")
plt.tight_layout(); plt.show()
# SVA cycles + average
gc = np.linspace(0,100,N_GC)
plt.figure(figsize=(8,4))
for c in sva_cycles:
    plt.plot(gc, c, alpha=0.35)
plt.plot(gc, sva_mean, lw=3, label="SVA mean")
plt.xlabel("Gait cycle (%)"); plt.ylabel("SVA (deg)"); plt.title("SVA
cycles (0-100% GC)"); plt.legend(); plt.tight_layout(); plt.show()
# EMG envelopes (averages)
if emg_means:
    plt.figure(figsize=(8,4))
    for ch, m in emg_means.items():
        plt.plot(gc, m, label=f"{ch} (env mean)")
    plt.xlabel("Gait cycle (%)"); plt.ylabel("EMG (a.u.)");
plt.title("EMG envelopes (cycle means)")
plt.legend(); plt.tight_layout(); plt.show()
# ===== EXPORT =====
# Save time-normalised means (SVA + EMG)
out = {"percent_gc": gc, "SVA_mean_deg": sva_mean}
for ch, m in emg_means.items(): out[f"{ch}_env_mean"] = m
pd.DataFrame(out).to_csv("/content/gc_norm_means.csv", index=False)
print("Wrote: /content/gc_norm_means.csv")

```