Toward a Personalized Knee–Ankle Wearable Exoskeleton for Partial Stroke Rehabilitation: A Review of Design, Control, and Clinical Evidence

Timothy Ayelagbe
Dept. of Electronic and Electrical Engineering
Obafemi Awolowo University
Ile-Ife, Nigeria
timothyayelagbe@gmail.com

Promise Osunyemi
Dept. of Electronic and Electrical Engineering
Obafemi Awolowo University
Ile-Ife, Nigeria
promiseosunyemi04@gmail.com

Glory Akanbi
Dept. of Electronic and Electrical Engineering
Obafemi Awolowo University
Ile-Ife, Nigeria
gloryinioluwa1@gmail.com

***Abstract*—Stroke remains a leading cause of long term physical disability worldwide, and a majority of survivors present with hemiparetic gait patterns marked by foot drop, knee hyperextension, reduced propulsion, and asymmetric stance. Wearable lower limb exoskeletons have emerged as a promising complement to conventional physiotherapy, yet most reported systems assist only a single joint, most commonly the ankle, and rely on generic rather than patient specific assistance profiles. This paper reviews recent clinical and technical literature on ankle and knee exoskeletons for post stroke gait rehabilitation, synthesizes the biomechanical rationale for combined knee and ankle assistance, and examines control architectures ranging from impedance based and adaptive oscillator schemes to biofeedback augmented assistance and variable-stiffness artificial-muscle actuation. Building on this synthesis, we outline an architecture for a personalized knee ankle exoskeleton intended for partial, task specific rehabilitation rather than full time ambulation support, and discuss design and deployment challenges relevant to resource constrained clinical settings such as those found in much of Sub Saharan Africa. The review is intended to guide future development of low cost, sensor driven, and individually tunable rehabilitation exoskeletons.**

***Index Terms*—Exoskeleton, stroke rehabilitation, hemiparetic gait, knee ankle orthosis, wearable robotics, biofeedback, gait control, adaptive frequency oscillator, variable stiffness actuation**

I. Introduction

Stroke is one of the leading causes of long term adult disability, and a large proportion of survivors experience hemiparesis, a unilateral weakness that disrupts the coordinated action of the hip, knee, and ankle during walking [1], [2]. Hemiparetic gait typically combines several deviations at once: impaired ankle dorsiflexion that produces foot drop during swing, excessive plantarflexion at heel strike, reduced push off propulsion, and knee hyperextension during stance that is used compensatorily to keep the paretic limb from buckling under load [3], [4]. Left unaddressed, chronic knee hyperextension increases joint loading and is associated with accelerated cartilage degeneration and secondary osteoarthritis [5], [6].

Passive ankle foot orthoses (AFOs) remain the most common clinical intervention. They stabilize the ankle and can improve toe clearance, but generally block active plantarflexion, restrict range of motion on inclines, and do little to correct knee hyperextension or spatiotemporal asymmetry [7]–[9]. These limitations have motivated a shift toward powered exoskeletons that can deliver active, tunable torque assistance rather than a fixed mechanical constraint [10].

The great majority of powered exoskeletons evaluated in post stroke populations assist the ankle alone, either through plantarflexion assistance to restore propulsion, dorsiflexion assistance to restore clearance, or both [12]–[14]. Far fewer devices jointly consider the knee, even though knee hyperextension is one of the most visible and clinically important compensations in hemiparetic gait, and only a small number of studies report knee kinematics alongside ankle assistance [15]. At the same time, dedicated knee exoskeletons and knee ankle foot orthoses developed for hemiparesis, such as the REFLEX platform, show that actively promoting a more normative knee trajectory can reduce reliance on compensatory strategies in the non paretic limb [16]. This split in the literature, ankle centric devices on one side and knee centric devices on the other, motivates a closer look at what a combined, personalized knee ankle exoskeleton would require.

This paper reviews recent experimental and survey literature on lower limb exoskeletons for stroke rehabilitation, with particular attention to (i) the biomechanical coupling between ankle and knee deviations in hemiparetic gait, (ii) control strategies used to assist the ankle and knee individually, including biofeedback augmented approaches and variable-stiffness bio-inspired actuation, and (iii) design considerations for extending single joint assistance into a coordinated, patient

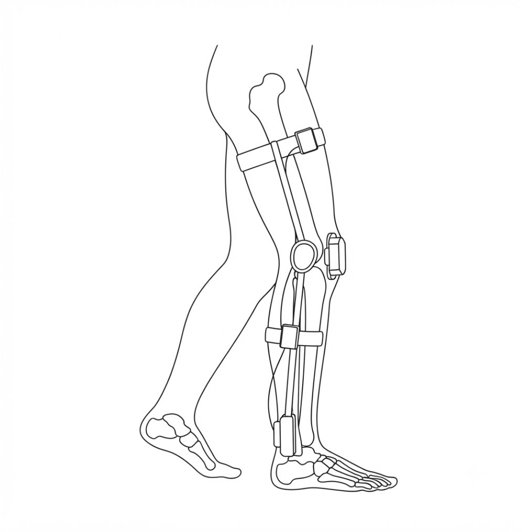

Fig. 1. Conceptual illustration of a personalized knee–ankle exoskeleton, showing thigh, shank, and foot cuffs linked by a joint mechanism spanning the knee and ankle, worn during gait. The layout reflects the actuation and cuff placement principles reviewed in Section IV: proximal thigh and distal shank/foot interfaces bracket a knee joint module and an ankle joint module, consistent with the series elastic, dual joint architecture discussed throughout this paper.

tunable knee ankle system. We further discuss the practical constraints of deploying such systems in low resource clinical environments, directly relevant to rehabilitation engineering in Nigeria and other Sub Saharan African settings, and propose a conceptual system architecture to guide future prototyping.

II. Biomechanics of Hemiparetic Gait

Understanding the biomechanical drivers of hemiparetic gait is a prerequisite for designing assistance that targets the right joint at the right phase of the gait cycle. Three interrelated deficits are consistently reported in the literature.

A. Ankle Deficits and Foot Clearance

Without assistance, the paretic ankle often remains plantarflexed through swing and at heel strike, producing foot drop and reduced minimum foot clearance, a recognized marker of increased fall risk [17]. Reported clearance gains from dorsiflexion assistance vary widely across devices, from a few millimeters to roughly two centimeters, with more recent autonomous designs demonstrating larger gains of three to five centimeters across a range of speeds and inclines [10], [15].

B. Knee Hyperextension

During early to mid stance, many hemiparetic individuals exhibit abnormally low knee flexion, effectively locking the knee to bear load safely. This hyperextension pattern is a compensatory response to weak or poorly timed ankle and quadriceps control, and it is only rarely corrected by ankle assistance alone [15]. Where ankle exoskeletons have restored near normative stance knee flexion, the effect appears to arise indirectly, through proper loading of the shank rather than direct knee actuation [15], suggesting a dedicated knee level assistance channel could produce more consistent, controllable outcomes.

C. Propulsion, Trailing Limb Angle, and Interlimb Symmetry

Self selected walking speed after stroke is strongly influenced by the propulsive force generated during late stance, commonly quantified as the anterior ground reaction force impulse [18], [19]. The trailing limb angle, the sagittal orientation of the limb relative to vertical at late stance, is highly correlated with this propulsive force and can be estimated in real time using shank and thigh mounted inertial sensors [20], [21].

A complementary line of evidence comes from unilateral knee-assistance studies in hemiparetic patients. The REFLEX knee–ankle–foot orthosis synchronizes assistance to the paretic knee with the phase of the nonparetic leg using an adaptive frequency oscillator (AO) [16]. Given the hip flexion angle of the unassisted leg $\theta_u(t)$, the AO estimates the continuous gait phase $\varphi_u(t)$ and frequency $\omega(t)$ through the coupled dynamic system

$$\varepsilon(t) = \theta_u(t) - \hat{\theta}_u(t), \tag{1}$$
$$\dot{\omega} = -\nu_\omega\, \varepsilon(t) \sin\varphi_u, \tag{2}$$
$$\dot{\varphi}_u = \omega - \nu_\varphi\, \varepsilon(t) \sin\varphi_u, \tag{3}$$
$$\hat{\theta}_u = \sum_{k=0}^{N_f} \alpha_k \cos(k\varphi_u) + \beta_k \sin(k\varphi_u), \tag{4}$$

where ν_ω, ν_φ are learning-rate constants and α_k, β_k are online-updated Fourier coefficients [16]. Because the two limbs are phase-shifted by approximately 180° in both healthy and hemiparetic gait [16], the assisted-leg phase is obtained as $\varphi_a = \hat{\varphi}_u + \pi$, tracking the user's own cadence rather than a fixed-frequency template. Reported RMS phase-estimation error for this scheme is below 2.4% and frequency error below 0.02 Hz [16], well within the phase-locking tolerance needed for synchronous multi-joint assistance. Clinically, REFLEX trials in three chronic stroke patients showed that synchronizing paretic-knee assistance to a normative healthy pattern, rather than mirroring the nonparetic limb, produced a more symmetric gait and reduced compensatory loading in the nonparetic leg, even though mirroring initially appeared to produce higher instantaneous symmetry [16]. This motivates using an AO-driven phase estimator as the timing backbone for a combined knee–ankle device, and favors a normative-pattern assistance law over simple contralateral mirroring.

Fig. 2 summarizes this sensing and actuation layout: inertial sensing on the nonparetic limb drives the phase estimate $\varphi(t)$, which in turn references the knee and ankle assistance delivered to the paretic limb.

III. State of the Art in Ankle and Knee Exoskeletons for Stroke

Table I summarizes the representative systems discussed in this section, alongside their joint coverage, actuation and sensing approach, control strategy, and key reported outcome.

TABLE I
REPRESENTATIVE KNEE AND/OR ANKLE EXOSKELETON SYSTEMS REVIEWED FOR POST-STROKE GAIT REHABILITATION

System / Study	Joint(s)	Actuation & Sensing	Control Strategy	Key Reported Outcome
Utah Ankle Exoskeleton [15]	Ankle	Series elastic actuator; onboard IMU/encoder	Impedance control at a neutral equilibrium angle with AO-timed plantarflexion assistance	Normalized heel-strike ankle angle; stance knee flexion rose from 7° to 20°; clearance up 3–5 cm
Speed-adaptive myoelectric ankle exo [13]	Ankle	EMG-driven series elastic actuator	Myoelectric proportional control scaled to gait speed	Increased paretic ankle power and walking speed; benefit sensitive to limb posture
Bilateral exosuit + vibrotactile biofeedback [21]	Ankle	Soft cable-driven exosuit; shank/thigh IMUs	Torque assistance plus real-time trailing-limb-angle biofeedback	Larger gains in propulsive impulse and speed than assistance or feedback alone
Cable-driven personalized exoskeleton [25]	Ankle (multi-joint capable)	Remote cable actuation; personalized profiles	Subject-tuned assistance magnitude and timing	Increased foot contact angle and knee flexion peak; reduced ankle flexor EMG
REFLEX knee-ankle-foot orthosis [16]	Knee	Motorized orthosis; AO on non-paretic limb	Stance-phase impedance reinforcement; swing-phase assisted-as-needed force tunnel	More symmetric gait and reduced non-paretic compensation with Pattern strategy
Rope-driven bionic knee exoskeleton [34]	Knee	Clutched aramid rope bundle, seven discrete stiffness levels	Task-dependent discrete stiffness selection	Reduced rectus femoris EMG; users preferred lower stiffness for level walking

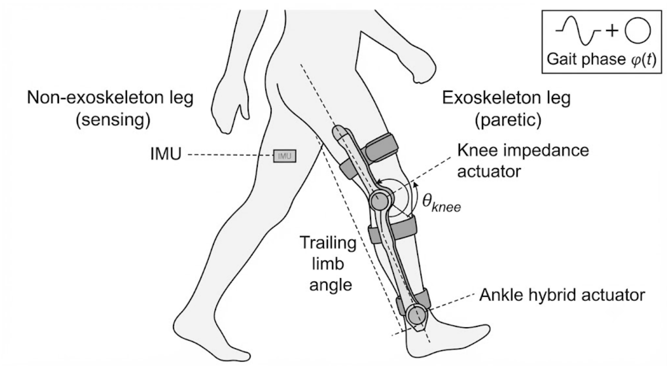

Fig. 2. Sensing and actuation layout for the proposed knee–ankle exoskeleton. IMUs mounted on the nonparetic (sensing) limb feed the AO-based gait-phase estimator of Eqs. (1)–(4), while the paretic (exoskeleton) limb carries the knee impedance actuator and ankle hybrid actuator, referenced respectively to the sagittal knee angle θ_{knee} and the trailing limb angle during stance.

A. Rigid Ankle Exoskeletons

Rigid, series elastic ankle exoskeletons have been used to deliver plantarflexion assistance during push off, dorsiflexion assistance during swing, or both. The Utah Ankle Exoskeleton combines an impedance controller that maintains a neutral equilibrium angle with an AO that times plantarflexion assistance to the gait cycle [15]. In a hemiparetic case study, this hybrid controller normalized ankle angle at heel strike, increased stance knee flexion from seven to twenty degrees, and increased minimum foot clearance by three to five centimeters across speeds and inclines, while reducing biological ankle effort [15]. Speed adaptive myoelectric controllers have similarly increased paretic ankle power and walking speed, although suboptimal limb posture can convert exoskeleton assistance into upward rather than forward propulsion [13]. Earlier neuromechanics based controllers demonstrated feasibility of powered ankle assistance in chronic stroke and reduced paretic soleus activation [12].

A second timing-adaptive design, ABLE-S, estimates stride time from bilateral shank kinematics and uses it to time pi-shaped plantarflexion (PF) and dorsiflexion (DF) torque profiles emulating unimpaired ankle dynamics, with safety layers that disable assistance if an invalid stride time or stop-walking state is detected [10]. Rather than foot-mounted force sensing, ABLE-S relies solely on shank angular position and velocity for gait event detection, argued to be more robust given the equinovarus deformity common after stroke [10]. In five chronic stroke participants, powered assistance increased paretic foot tilting angle at heel strike by 349.8% relative to no exoskeleton, increased foot clearance by 24.6%, increased peak step length by 47.0%, and reduced paretic hip hiking by 55.9% relative to the unpowered condition [10]. Hip hiking fell relative to unpowered but not relative to no-exoskeleton, attributed by the authors to added distal mass, underscoring that ankle-only assistance does not reliably correct compensatory knee and hip patterns [10]. The device weighs 2.80 kg total and required on average 7.86 ± 2.90 minutes to don, doff, and tune, with manual parameter tuning alone consuming 43% of that time [10] – a usability bottleneck directly relevant to the personalization layer discussed in Section IV-D.

B. Soft and Cable Driven Exosuits

Textile based exosuits reduce distal mass and increase self-selected and maximum walking speed in stroke survivors, though metabolic benefit has been less consistent than for unimpaired users [22], [23]. Benefit depends strongly on baseline walking ability, with limited improvement in more severely impaired individuals [24]. Cable driven architectures extend this to multi joint assistance with remote actuation; a hybrid cable driven exoskeleton with personalized profiles increased foot contact angle and knee flexion peak while reducing ankle flexor activity in subacute stroke, illustrating the value of individualized assistance [25].

C. Biofeedback Augmented Assistance

Because exoskeleton benefit depends on how well the user positions the limb, several groups pair robotic assistance with real time biofeedback. Audiovisual feedback on propulsive force or trailing limb angle improves propulsion in able bodied and post stroke populations [26]–[28]. A wearable vibrotactile-audio biofeedback system combined with bilateral ankle exoskeletons produced greater increases in paretic propulsive impulse and walking speed than either intervention alone [21]. Reviews of augmented feedback in motor learning support the principle that feedback must be simple, timely, and specific to a single, clinically meaningful variable [29], [30].

D. Knee Focused Devices and Bio-Inspired Variable Stiffness

Dedicated knee exoskeletons remain comparatively rare. REFLEX uses an AO on the nonparetic limb to estimate gait phase, following Eqs. (1)–(4), and assists the paretic knee by mirroring the healthy limb ("Echo") or driving it toward a normative reference trajectory ("Pattern") [16]. Targeting a healthy reference pattern produced better long term limb behavior than mirroring, despite an initial reduction in interlimb symmetry [16]. The device applies a stance-phase high-impedance "limb reinforcement" law and a swing-phase assisted-as-needed force-tunnel law, switched by insole-detected gait events [16], a pattern directly reusable for a combined knee channel.

A separate direction targets joint impedance directly. Jin et al. present a bionic knee exoskeleton actuated by rope-driven artificial muscles: a clutch engages one to four aramid-fiber ropes of graded diameter, producing seven discrete stiffness levels from one actuator [34]. The combined stiffness is

$$K = K_1 + K_2 + \cdots + K_n, \qquad K_i = \frac{M_i k_i}{l_i + \Delta l_i}, \tag{5}$$

where M_i is the rope's breaking force, k_i its dynamic stiffness coefficient, and l_i, Δl_i its length and elongation [34]. Wearers preferred low stiffness for level walking and higher stiffness for running and incline ascent, with reduced rectus femoris EMG activation indicating genuine load sharing [34] – a low-part-count mechanism attractive when embedded computation is limited, as in a low-cost field-deployable device.

E. Broader Surveys and Meta Analyses

Systematic reviews of wearable lower limb exoskeletons catalogue mechanical structures, actuation schemes, and intent recognition methods, highlighting bionic structure and controller adaptability as recurring open problems [31], [32], [35]. Hasan and Alam's comparative analysis in particular contrasts control, design, and application choices across recent devices, reinforcing that actuation topology and control law are typically optimized jointly rather than independently [35]. A meta analysis across six randomized controlled trials found hip assistance more effective for step length, cadence, and symmetry, and ankle assistance more effective for walk test performance, suggesting the optimal joint to assist may depend on the patient's impairment profile [33], supporting a device that can flexibly allocate assistance between knee and ankle.

A large systematic review of 159 lower-limb exoskeleton studies across 43 devices for brain injury rehabilitation provides quantitative support for several choices above [11]. Only 10.5% of stroke-validated devices implement challenge-based (resistive) control, and just 14% adapt assistive parameters automatically in real time. Threshold-based synchronization dominates the field (72.1% overall, most commonly ground-reaction-force thresholds from foot-mounted sensors, 62.8%); AO-based synchronization, the family the REFLEX and proposed estimators rely on, was used in only 10.5% of stroke exoskeletons, while kinematic synchronization was used in 41.8%, argued to be more robust for the irregular center-of-pressure trajectories of hemiplegic gait [11]. Critically, for chronic stroke, assistive control combining AO synchronization with trajectory-tracking mid-level control achieved the strongest evidence-graded improvements with the lowest training intensity among twelve compared strategy combinations – 46.0% improvement in the 6-Minute Walk Test and 34.0% in the 10-Meter Walk Test [11], directly supporting anchoring the proposed architecture's phase estimator to an AO formulation rather than a simpler threshold scheme.

IV. Design Considerations for a Combined Knee Ankle System

Synthesizing the reviewed literature, four design considerations recur across successful single joint devices and are directly relevant to a combined knee ankle system.

A. Actuation and Structure

Series elastic actuation with a low level two degree of freedom torque tracking loop has proven effective for both joints, since spring deflection gives a low noise interaction-torque estimate without a dedicated load cell [15]. For a combined device, decoupling knee and ankle actuators while sharing a single battery and control unit, as in cable driven designs, reduces distal mass without sacrificing torque capability [25]. Discretely-switched variable-stiffness rope-bundle actuation [34] offers a mechanically simple way to trade compliance for load-bearing capacity between stance-reinforcement and swing-guidance phases, mirroring the impedance-switching logic already validated in REFLEX [16].

B. Sensing and Gait Phase Estimation

Thigh and shank mounted IMUs allow real time estimation of joint angles and derived variables such as trailing limb angle, with reported correlation above 0.99 against motion capture [21], as illustrated in Fig. 2. IMUs on the nonparetic limb feeding an AO as in Eqs. (1)–(4) have independently been validated for knee-phase synchronization with sub-2.4% RMS phase error [16]. This is attractive for a combined device because the same sensor set supports both ankle phase estimation and knee hyperextension detection, avoiding additional instrumentation, and the AO's phase correction at each heel strike keeps the estimate anchored to true gait events

[16]. It is also consistent with the systematic-review finding that kinematic synchronization is comparatively under-used (41.8%) relative to ground-reaction-force thresholds (62.8%) despite its robustness advantage [11], and avoids the shoe-embedded force sensing that ABLE-S relies on for its own detection [10].

C. Control Strategy

The most robust controllers combine an impedance term, enforcing a safe reference posture via stiffness and damping, with an event or oscillator driven torque term timing active assistance to specific phases [15], [16]. We argue the knee channel should primarily use an impedance term tuned to prevent hyperextension during early stance, while the ankle channel uses a torque term timed to late stance push off and early swing dorsiflexion, coordinated through a shared finite state machine driven by the estimated gait phase, in the spirit of REFLEX's stance-reinforcement/swing-guidance law [16]. This is further supported at the taxonomy level: across 159 reviewed studies, controllers combining trajectory-tracking with compliant control, synchronized via an AO rather than a fixed threshold, achieved the highest reported clinical improvements with the lowest training intensity for chronic stroke [11].

D. Personalization and Biofeedback

Assistance hand tuned per participant, rather than fixed across users, is repeatedly associated with better outcomes [13], [25]. REFLEX's Pattern strategy extracts the nonparetic leg's per-step range of motion and phase-of-peak-flexion, low-pass filters these over a five-step buffer, and uses them to scale and time-shift a normative trajectory applied to the paretic limb [16]. ABLE-S offers a lower-tech example: PF/DF torque magnitudes were manually tuned per participant to concrete clinical targets (5–10° dorsiflexion at initial contact; 5–20% of unimpaired peak PF torque) [10], but this manual tuning consumed the largest single share (43%) of total setup time [10], and only 14% of reviewed exoskeletons implement any automatic real-time adaptation [11] – together motivating the buffered, automatic feature-extraction approach of REFLEX over purely manual tuning here. Layering biofeedback on top of mechanical assistance further improves the odds that the user adopts the posture needed to receive that assistance effectively [21]. A personalized system should therefore expose a small number of clinically interpretable tuning parameters, such as target trailing limb angle and target stance knee flexion, rather than requiring raw torque-profile tuning.

V. Proposed Conceptual Architecture

Fig. 3 summarizes a conceptual architecture for a personalized knee ankle exoskeleton intended for partial, task specific rehabilitation sessions rather than continuous ambulation support. The layout follows a single left-to-right processing pipeline of five stages – sensing, gait-phase estimation, personalization, dual-joint control, and actuation – with session logging and biofeedback drawn as outputs, and torque-tracking feedback routed along a single dedicated return path beneath the pipeline so that no arrow re-crosses into an upstream stage. Fig. 1 shows the corresponding physical layout of the wearable device itself, with thigh, shank, and foot cuffs bracketing knee and ankle joint modules.

Thigh and shank mounted inertial sensors on the nonparetic limb feed a gait phase estimator implementing the AO of Eqs. (1)–(4), which classifies the current portion of the gait cycle and computes trailing limb angle and knee flexion angle in real time, following Herrin et al. [21] and the AO formalism of Lora-Millan et al. [16]. We favor this AO formulation over a ground-reaction-force threshold scheme (used in 72.1% of existing devices [11], including ABLE-S's own shank-kinematics detector [10]) because it is the synchronization method associated with the strongest chronic-stroke outcomes in the largest available cross-device comparison [11]. The nonparetic-leg phase is shifted by π rad to obtain the paretic-leg reference phase φ_a, exactly as in REFLEX [16]. These estimates pass to two parallel controllers: a knee impedance controller resisting hyperextension during early to mid stance, and an ankle hybrid torque-impedance controller assisting dorsiflexion during swing and plantarflexion during push off, in the style of the Utah controller [15]. A personalization layer upstream of both controllers stores per-patient targets obtained via REFLEX's buffered feature-extraction method [16] and adjusts controller gains accordingly, echoing tuning procedures from cable driven and myoelectric devices [13], [25] and the clinical PF/DF torque targets validated for ABLE-S [10]. A vibrotactile-auditory biofeedback module runs alongside mechanical assistance, cueing the patient toward the target trailing limb angle in real time [21]. Low level torque tracking at each joint uses a two degree of freedom closed loop controller referenced to series elastic spring deflection [15], optionally paired at the knee with the discrete rope-bundle actuator of Jin et al. [34] using Eq. (5) to set gait-mode-appropriate impedance, while a supervisory layer logs gait metrics across sessions for clinician review – a metric reported by remarkably few existing studies despite its practical importance [11].

VI. Challenges and Future Directions

Several open challenges remain before a combined knee ankle exoskeleton of this kind can move from concept to clinical use.

A. Coordinated Multi Joint Control

Most reported controllers were designed and validated for a single joint. Extending an impedance and oscillator based scheme to two coupled joints risks conflicting torque commands, e.g. when knee stabilization and ankle push off assistance overlap in early to mid stance, requiring joint level controllers explicitly coordinated through a shared gait phase estimate rather than tuned independently [15], [16].

B. Evidence from Larger and More Diverse Cohorts

The strongest evidence for both ankle and knee assistance currently comes from case studies and small feasibility trials

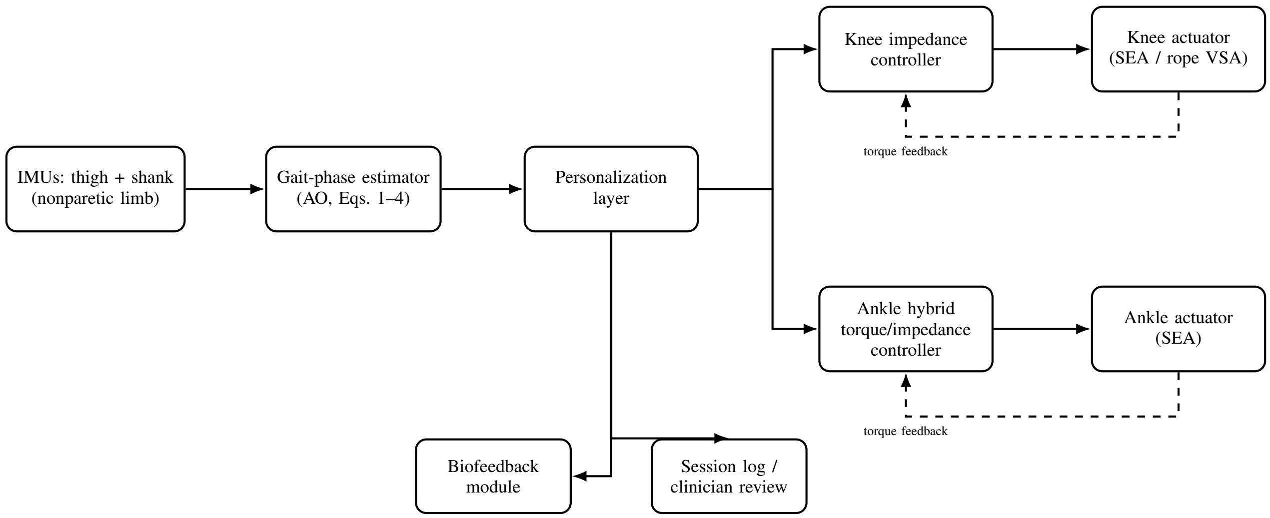

Fig. 3. Proposed conceptual architecture of the personalized knee–ankle exoskeleton, drawn as a single left-to-right pipeline. Solid arrows show the forward sensing-to-actuation path; dashed arrows show the only feedback in the system, local torque tracking at each joint. Session logging and biofeedback are outputs of the personalization stage rather than being folded into the forward path, so no arrow re-crosses an upstream stage.

of one to seven participants [10], [15], [16], [21], [25], [34]. Larger, more heterogeneous cohorts are needed to confirm generalization across stroke severity, time since stroke, and gait pattern [33]. This is corroborated field-wide: across the 57 stroke studies compared in [11], mean participants per study was only 14.87 ± 13.53, mean training sessions 11.77 ± 12.20, and 66.04% of studies were observational rather than controlled trials.

C. Personalized Tuning at Scale

Hand tuning by an experienced clinician yields good outcomes in laboratory studies but is difficult to scale to routine clinical use [13], [25]. In ABLE-S, manual tuning was the single largest contributor to setup time (3.39 ± 1.36 of 7.86 ± 2.90 minutes total), exceeding even donning time [10], and only 14% of reviewed exoskeletons implement any automatic real-time parameter adaptation [11]. Automated tuning informed by estimated trailing limb angle and knee flexion trajectory, using REFLEX's buffered feature-extraction approach combined with biofeedback, could reduce clinician time per session [16], [21].

D. Cost, Portability, and Deployment in Resource Constrained Settings

Most reviewed devices were developed in well resourced motion capture laboratories and rely on relatively expensive series elastic actuation and embedded computing. For deployment in Nigeria and other Sub Saharan African contexts, where access to specialized rehabilitation facilities and trained physiotherapists is limited, a personalized knee ankle exoskeleton should prioritize low cost inertial sensing over motion capture, minimal onboard computation, and parameters settable from a small number of short calibration trials rather than extended laboratory sessions. The discrete rope-bundle actuator of Jin et al. [34] is a promising low-cost option, replacing continuously variable impedance electronics with a small servo-driven clutch. Likewise, a shank-kinematics-only detector in the style of ABLE-S [10] avoids the shoe-embedded force sensing that dominates the wider field (62.8% of devices [11]), simplifying footwear compatibility and lowering instrumentation cost. This area is largely absent from the current literature and represents a concrete direction for future locally driven research.

E. Safety, Usability, and Regulatory Pathways

Because the device described here targets supervised, task-specific rehabilitation sessions rather than unsupervised ambulation, its safety case is comparatively tractable: impedance-dominant knee control limits its peak torque even under sensor faults, and session-bounded operation reduces exposure relative to all-day assistive use. Even so, translating a laboratory-validated controller into a clinic-ready device requires addressing donning and doffing time for patients with limited dexterity, skin interface tolerance over repeated sessions, fail-safe behavior on IMU dropout, and eventual alignment with applicable medical device regulatory pathways – considerations rarely reported in the literature reviewed here.

VII. Conclusion

This review has examined recent ankle and knee exoskeleton research for post stroke gait rehabilitation, drawing on autonomous ankle exoskeletons, biofeedback augmented assistance, cable driven personalized exosuits, an AO-driven knee orthosis (REFLEX), a rope-driven variable-stiffness bionic knee exoskeleton, a timing-adaptive ankle exoskeleton with quantified outcomes (ABLE-S), and a 159-study systematic review of control strategies across the field. Three themes stand out: hemiparetic gait deviations at the ankle and knee are mechanically coupled and not fully addressed by single joint assistance; phase aware controllers combining an impedance term with a timed torque term consistently outperform fixed profile assistance, and AO synchronization with trajectory-tracking control shows the strongest cross-study evidence for chronic stroke despite being used in only a small minority of devices [11]; and personalization, whether by hand tuning, buffered feature-based scaling, or biofeedback, is a recurring driver of clinically meaningful improvement, even though manual tuning remains a disproportionate share of setup time in practice [10]. Building on this synthesis, we proposed a conceptual, non-overlapping architecture for a personalized knee ankle exoskeleton that coordinates dual joint assistance through a shared AO gait phase estimate and layers biofeedback and patient specific tuning atop a series elastic, optionally variable-stiffness, actuation base. Realizing this architecture in a low cost, field deployable form remains an open and clinically important engineering problem, particularly for rehabilitation settings with limited access to specialized equipment.

References

[1] E. J. Benjamin *et al.*, "Heart disease and stroke statistics, 2019 update," *Circulation*, vol. 139, no. 10, pp. e56–e528, Mar. 2019.

[2] M. M. Rybar *et al.*, "The stroke related effects of hip flexion fatigue on overground walking," *Gait Posture*, vol. 39, no. 4, pp. 1103–1108, 2014.

[3] D. Mozaffarian *et al.*, "Heart disease and stroke statistics, 2016 update," *Circulation*, vol. 133, no. 4, pp. e38–360, 2016.

[4] M. Appasamy *et al.*, "Treatment strategies for genu recurvatum in adult patients with hemiparesis: A case series," *PM&R*, vol. 7, no. 2, pp. 105–112, Feb. 2015.

[5] S. Korkusuz *et al.*, "Effect of knee hyperextension on femoral cartilage thickness in stroke patients," *Am. J. Phys. Med. Rehabil.*, vol. 103, no. 5, pp. 371–376, 2023.

[6] S. Hada *et al.*, "The degeneration and destruction of femoral articular cartilage in early stage knee osteoarthritis," *Osteoarthr. Cartil.*, vol. 22, no. 10, pp. 1583–1589, 2014.

[7] K. Pongpipatpaiboon *et al.*, "The impact of ankle foot orthoses on toe clearance strategy in hemiparetic gait," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, p. 41, 2018.

[8] A. Vistamehr, S. A. Kautz, and R. R. Neptune, "The influence of solid ankle foot orthoses on forward propulsion and dynamic balance in healthy adults during walking," *Clin. Biomech.*, vol. 29, no. 5, pp. 583–589, 2014.

[9] T. H. Cruz and Y. Y. Dhaher, "Impact of ankle foot orthosis on frontal plane behaviors post stroke," *Gait Posture*, vol. 30, no. 3, pp. 312–316, 2009.

[10] J. de Miguel-Fernandez, C. Pescatore, A. Mesa-Garrido, C. Rikhof, E. Prinsen, J. M. Font-Llagunes, and J. Lobo-Prat, "Immediate biomechanical effects of providing adaptive assistance with an ankle exoskeleton in individuals after stroke," *IEEE Robot. Autom. Lett.*, vol. 7, no. 3, pp. 7574–7580, Jul. 2022.

[11] J. de Miguel-Fernandez, J. Lobo-Prat, E. Prinsen, J. M. Font-Llagunes, and L. Marchal-Crespo, "Control strategies used in lower limb exoskeletons for gait rehabilitation after brain injury: a systematic review and analysis of clinical effectiveness," *J. Neuroeng. Rehabil.*, vol. 20, art. 23, Feb. 2023.

[12] K. Z. Takahashi, M. D. Lewek, and G. S. Sawicki, "A neuromechanics based powered ankle exoskeleton to assist walking post stroke," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, p. 23, 2015.

[13] E. M. McCain *et al.*, "Mechanics and energetics of post stroke walking aided by a powered ankle exoskeleton with speed adaptive myoelectric control," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, p. 57, 2019.

[14] C. Siviy *et al.*, "Offline assistance optimization of a soft exosuit for augmenting ankle power of stroke survivors during walking," *IEEE Robot. Autom. Lett.*, vol. 5, no. 2, pp. 828–835, 2020.

[15] K. Pruyn, R. Murray, L. Gabert, and T. Lenzi, "Autonomous powered ankle exoskeleton improves foot clearance and knee hyperextension after stroke: A case study," *IEEE Trans. Med. Robot. Bionics*, vol. 7, no. 1, pp. 51–58, 2024.

[16] J. S. Lora-Millan, F. J. Sanchez-Cuesta, J. P. Romero, J. C. Moreno, and E. Rocon, "A unilateral robotic knee exoskeleton to assess the role of natural gait assistance in hemiparetic patients," *J. Neuroeng. Rehabil.*, vol. 19, art. 109, 2022.

[17] H. Nagano, C. M. Said, L. James, W. A. Sparrow, and R. Begg, "Biomechanical correlates of falls risk in gait impaired stroke survivors," *Front. Physiol.*, vol. 13, art. 833417, 2022.

[18] S. A. Roelker, M. G. Bowden, S. A. Kautz, and R. R. Neptune, "Paretic propulsion as a measure of walking performance and functional motor recovery post stroke," *Gait Posture*, vol. 68, pp. 6–14, 2019.

[19] M. G. Bowden, C. K. Balasubramanian, R. R. Neptune, and S. A. Kautz, "Anterior posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking," *Stroke*, vol. 37, no. 3, pp. 872–876, 2006.

[20] M. D. Lewek and G. S. Sawicki, "Trailing limb angle is a surrogate for propulsive limb forces during walking post stroke," *Clin. Biomech.*, vol. 67, pp. 115–118, 2019.

[21] K. R. Herrin, Y. T. Pan, T. M. Kesar, G. S. Sawicki, and A. J. Young, "Robotic ankle exoskeleton and limb angle biofeedback for assisting stroke gait," *IEEE Robot. Autom. Lett.*, vol. 10, no. 2, pp. 1011–1017, 2024.

[22] L. N. Awad *et al.*, "A soft robotic exosuit improves walking in patients after stroke," *Sci. Transl. Med.*, vol. 9, no. 400, art. eaai9084, 2017.

[23] L. N. Awad, P. Kudzia, D. A. Revi, T. D. Ellis, and C. J. Walsh, "Walking faster and farther with a soft robotic exosuit," *IEEE Open J. Eng. Med. Biol.*, vol. 1, pp. 108–115, 2020.

[24] L. H. Sloot *et al.*, "Effects of a soft robotic exosuit on the quality and speed of overground walking depends on walking ability after stroke," *J. Neuroeng. Rehabil.*, vol. 20, no. 1, p. 113, 2023.

[25] B. Zhong *et al.*, "A cable driven exoskeleton with personalized assistance improves the gait metrics of people in subacute stroke," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 31, pp. 2415–2426, 2023.

[26] J. Liu, H. B. Kim, S. L. Wolf, and T. M. Kesar, "Comparison of the immediate effects of audio, visual, or audiovisual gait biofeedback on propulsive force generation," *Appl. Psychophysiol. Biofeedback*, vol. 45, pp. 211–220, 2020.

[27] J. Liu, V. Santucci, S. Eicholtz, and T. M. Kesar, "Comparison of the effects of real time propulsive force versus limb angle gait biofeedback on gait biomechanics," *Gait Posture*, vol. 83, pp. 107–113, 2021.

[28] K. Genthe *et al.*, "Effects of real time gait biofeedback on paretic propulsion and gait biomechanics in individuals post stroke," *Top. Stroke Rehabil.*, vol. 25, no. 3, pp. 186–193, 2018.

[29] R. Sigrist, G. Rauter, R. Riener, and P. Wolf, "Augmented visual, auditory, haptic, and multimodal feedback in motor learning: A review," *Psychon. Bull. Rev.*, vol. 20, pp. 21–53, 2013.

[30] M. R. Afzal *et al.*, "Effects of vibrotactile biofeedback coding schemes on gait symmetry training of individuals with stroke," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 27, no. 8, pp. 1617–1625, 2019.

[31] S. Qiu, Z. Pei, C. Wang, and Z. Tang, "Systematic review on wearable lower extremity robotic exoskeletons for assisted locomotion," *J. Bionic Eng.*, vol. 20, pp. 436–469, 2023.

[32] A. Foroutannia and M. Mohammadian, "A comprehensive survey of lower limb assistive exoskeleton robots," *Robot. Auton. Syst.*, vol. 195, art. 105232, 2026.

[33] J. Liu, K. Liu, X. Ge, P. Zhou, T. Bao, and W. Gong, "Flexible exoskeleton assisted training enhances lower limb motor function after stroke," *J. Neurol.*, 2025.

[34] S. Jin, B. Liu, and Z. Wang, "A bionic knee exoskeleton design with variable stiffness via rope-based artificial muscle actuation," *Biomimetics*, vol. 10, no. 7, art. 424, 2025.

[35] S. Hasan and N. Alam, "Comprehensive comparative analysis of lower limb exoskeleton research," *Actuators*, vol. 14, no. 7, art. 342, 2025.