DRBA: Dynamic Robotic Balance Assistant - An assist-as-needed gait and balance rehabilitation robot for versatile training

Yifan Wang*[1], Lei Li*[2], Youlong Wang[1], Chengyuan Yang[1], Sherwin Stephen Chan[1], Jiaye Chen[1],
Hao Wang[2], Xuesheng Gong[3], Jun Lin[3], Hongping Hu[3], and Wei Tech Ang[1]

Abstract— **The decline of human balance control due to aging and pathological conditions increases fall risk, a major concern in geriatric care and rehabilitation. Gait training is essential for balance recovery, enhancing walking ability and postural control. However, existing overground robotic gait trainers have limitations: body weight support systems are bulky and impractical for daily use, while end-effector-based systems often compromise transparency, altering natural gait dynamics. This paper presents the Dynamic Robotic Balance Assistant (DRBA), a novel gait trainer providing assist-as-needed body weight and balance support for various training scenarios. DRBA integrates a 3-degree-of-freedom (3-DoF) robotic arm for pelvic support with flexible motion, a compact sit-to-stand assistance module, and user-following and fall detection algorithms to ensure minimal interference and responsive support. Experimental results demonstrated high transparency, with minimal impact on natural gait dynamics. A patient trial with nine elderly patients with varying medical conditions and balance impairments (ranging from severe to mild) further validated DRBA's effectiveness. The results showed that DRBA-assisted training increased step length and walking speed compared to therapist-assisted gait training. Additionally, DRBA enabled users to perform tasks beyond their unaided ability, expanding rehabilitation possibilities. These findings highlight DRBA's potential to enhance rehabilitation outcomes by facilitating higher training intensity and enabling task-oriented exercises.**

Index Terms— **Rehabilitation Robotics, Physical human-robot interaction, Compliant Joints and Mechanisms**

I. INTRODUCTION

Human balance control naturally declines with age and is further affected by neurological disorders (e.g., stroke, spinal cord injury, Parkinson's disease), musculoskeletal conditions (e.g., chronic low back pain, scoliosis, amputation), and vestibular deficits (e.g., benign paroxysmal positional vertigo). These impairments often lead to reduced proprioception and coordination, increasing the risk of falls, which poses significant challenges for elderly and impaired individuals undergoing rehabilitation [1]. Given that balance control is critical for activities of daily living (ADL) [2] and is a strong predictor of independent living [3], its deterioration significantly impacts both physical function and quality of life. Additionally, impaired balance is a leading cause of self-perceived disability post-rehabilitation [4], underscoring the need for effective interventions to restore balance and mobility.

Gait training is a widely recognized approach for balance recovery [5], [6], [7], as it improves both walking ability and postural control by integrating balance mechanisms into locomotion. Over the past decade, overground gait and balance trainers have been developed to enhance rehabilitation outcomes, reduce caregiver burden, and improve safety during gait training

Overground gait trainers generally fall into two categories: body weight support systems and end-effector-based systems. Body weight support systems, such as Andago [8], may include a mobile base [8], [9] or function without one [10], [11]. These systems provide continuous partial or full body weight support through a suspended harness mechanism, reducing physical strain on the patient during rehabilitation. However, their bulky structural design and large footprint limit maneuverability, requiring spacious training environments, making them primarily suitable for hospital and rehabilitation institution use.

End-effector-based systems, such as KineAssist [12], integrate a robotic arm for user interaction coupled with a mobile base that moves with the user. Compared to body weight support systems, these robots are more compact and flexible, making them suitable for outdoor and community-based rehabilitation. However, existing end-effector-based systems often struggle with transparency issues in their physcial human robot interface (pHRI) design, imposing unintended mechanical constraints on the user's natural movement, ultimately reducing gait training effectiveness. Early robotic gait trainers, such as KineAssist [12] and SoloWalk [13], utilized rigid robotic arms for support and protection. However, these rigid structures transmitted inertial forces from the mobile base to the user, distorting gait dynamics and altering natural gait patterns. Later systems attempted to mitigate these inertial effects using smooth tracking controllers based on human-robot interaction forces [14], [15], but gait alterations were still observed [16]. Additionally, their rigid mechanical constraints restricted user movement, limiting engagement in diverse rehabilitation tasks and ADL activities.

Recent efforts have introduced compact robotic gait trainers designed for home and community use [17], [18]. These systems integrate passive parallel robotic arms to decouple mobile base dynamics from the user, improving gait training transparency. However, their integration with electrically powered wheelchairs, intended for daily transportation, com-

---

*indicates equal contribution

[1] Yifan Wang, Youlong Wang, Chengyuan Yang, Sherwin Stephen Chan, Jiaye Chen and Wei Tech Ang are with School of Mechanical and Aerospace Engineering, Nanyang Technological University, 639798 Singapore `Corresonding email: ywang114@e.ntu.edu.sg`

[2] Lei Li, and Hao Wang are with Guangdong Zhongxin Intelligent Rehabilitation Research Institute, Foshan, China 528200

[3] Xuesheng Gong, Jun Lin and Hongping Hu are with Guangdong Jianxiang Hospital Group, Foshan, China 528200

promises pHRI transparency by restricting free movement. Additionally, the large inertia of the mobile base induces a pulling-pushing effect, further disrupting natural gait patterns during training [17].

Despite advancements, existing robotic gait trainers still face trade-offs among mobility, transparency, and adaptability. Body weight support systems, though effective, are bulky and limited to clinical settings, while end-effector-based systems constrain natural gait. Even newer compact designs restrict free movement and introduce unwanted dynamics. These challenges underscore the need for a lightweight, adaptable, and transparent gait trainer that preserves natural movement across diverse rehabilitation environments.

In this paper, we introduce **DRBA** – Dynamic Robotic Balance Assistant, a novel gait rehabilitation robot designed to provide assist-as-needed body weight and balance support for level-ground training. DRBA is engineered for versatile deployment in rehabilitation centers, community settings, and outdoor environments. It features an innovative balance assistance mechanism with a 3-degree-of-freedom (3-DoF) robotic arm that actively monitors the user's state and provides adaptive pelvic support when needed. This design preserves natural pelvic motion, enabling users to engage in functional and challenge-based rehabilitation tasks. A smooth user-following algorithm and sensitive fall detection system ensure minimal interference when assistance is not required while providing timely and effective balance support during instability. To facilitate long-distance mobility, DRBA incorporates a compact sit-to-stand assistance system for seamless transitions between sitting and standing.

Experimental results demonstrated that DRBA maintains high transparency, as indicated by minimal alterations to users' gait patterns and stability. Pilot studies involving nine patients with Berg Balance Scale (BBS) scores ranging from 8 to 50 and various medical conditions confirmed its effectiveness. Gait analysis revealed that DRBA-assisted training improved step length and walking speed compared to therapist-assisted training while also enhancing users' perceived safety. Furthermore, DRBA enabled users to perform tasks beyond their unaided physical capacity, expanding rehabilitation options and training diversity. These findings suggest that DRBA can enhance rehabilitation outcomes by facilitating higher training intensity through adaptive assistance, promoting more frequent and prolonged rehabilitation sessions (frequency and duration), and enabling task-oriented exercises tailored to individual patient needs, following the FITT principle [19].

Our key contributions include:

- **Dynamic Robotic Balance Assistant (DRBA):** A novel robotic gait trainer that provides assist-as-needed body weight and balance support, preserving natural pelvic motion within a compact and mobile form factor, making it suitable for rehabilitation centers, community care settings, and outdoor use.
- **Patient Trials:** We conducted patient studies with nine patients with various medical conditions and balance impairments (BBS: 8–50). Results demonstrated significant improvements in gait parameters and enhanced safety, enabling higher-intensity exercises and a broader range of rehabilitation activities.

II. SYSTEM OVERVIEW

DRBA comprises three subsystems: 1) Balance assistance system, 2) Sit-to-stand assistance system and 3) Mobility assistance system to assist and enhance users' balance and mobility training. An overview of the DRBA system is presented in Fig. 1. The control architecture integrates a user-following controller and an instability detection and fall intervention mechanism, enabling the robot to synchronously accompany the user and provide timely balance support when instability is detected.

A. Balance assistance system

The balance assistance system continuously monitors the user's state and provides support as needed. It features a low-inertia compliant robotic arm that wraps around the user's pelvis. Similar to systems like KineAssist [12] and SoloWalk [13], this design targets the pelvis—located near the center of mass (CoM)—to deliver balance interventions more effectively. However, unlike these systems, which use rigid robotic arms, our design incorporates a passive, compliant structure that allows 3-DoF planar relative motion of the user's pelvis. This critical design choice decouples the robot's main body inertia from the user during overground gait training, minimizing impedance forces when balance assistance is not required. The robotic arm primarily enables pelvic motion in the transverse plane, but our pHRI design incorporates elastic cushions and adjustable belts to provide a unique yet limited degree of freedom in other directions, such as pelvic tilt, which depends on the deformation of these materials. This flexible design enables users to perform functional and ADL tasks, such as kicking a ball or wearing pants, which require vertical CoM manipulation.

The robotic arm consists of six rotary joints with a pelvis interface. Four of these joints are equipped with encoders and brakes. The encoders measure the position and orientation of the interface, enabling continuous monitoring of the relative position between the user and the robot. The brakes provide balance assistance and fall intervention by locking the robotic arm to stabilize the user in times of instability. Additionally, two force sensors are symmetrically mounted on the pelvis interface to detect the user's body weight, which is transferred through the safety belt. Additionally, detachable handles are attached to the pelvis interface, providing users with limited mobility an optional self-support mechanism as needed.

B. Sit-to-stand assistance system

The sit-to-stand assistance system enables safe, seamless transfers from hospital beds to rehabilitation gyms, reducing therapist workload. It features a vertical linear slider powered by a linear driving module. A foldable seat is attached to both the slider and the base of the robot. The seat's linkage joint moves within a sliding groove on the base, which has an

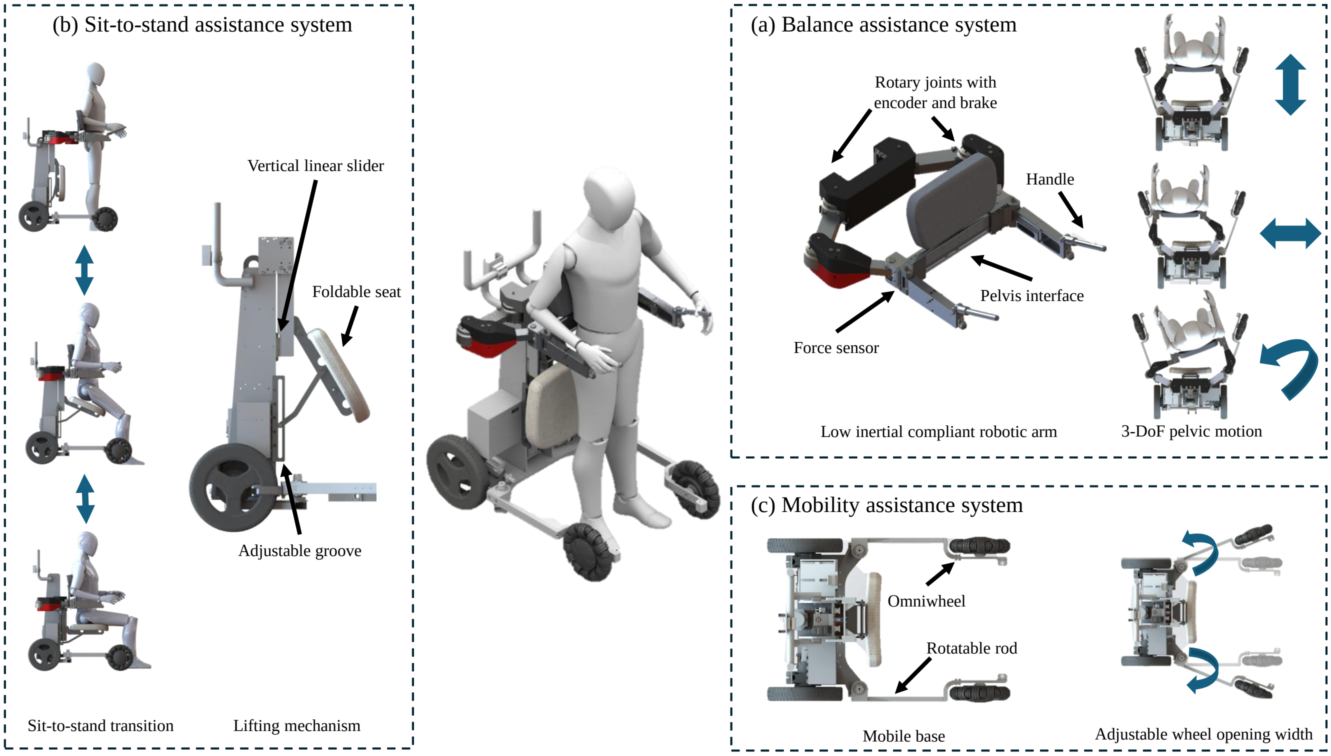

Fig. 1. System Overview of DRBA. (a) Balance assistance system: monitors the user's state and provides adaptive balance support through a low inertia compliant robotic arm, enabling 3-DoF pelvic motion. (b) Sit-to-stand assistance system: assists users in transitioning between sitting and standing, featuring a lifting mechanism with a vertical linear slider and a foldable seat. (c) Mobility assistance system: allows the robot to move together with the user via a mobile base with two omniwheels mounted on rotatable rods, enabling adjustable wheel spacing to accommodate different walking needs.

adjustable limit to accommodate different seat heights based on user needs. The transition from sitting to standing occurs in two stages: i) folding stage: the seat linkage operates as a four-bar linkage, shifting the seat from a horizontal to a vertical position as it folds onto the lifting mechanism while the slider ascends. ii) rising stage: once the seat reaches the vertical position, it remains upright and continues to move with the slider until the joint in the groove reaches its top limit. Separately, the 3 DoF robotic arm is also mounted on the slider which allows the user to adjust an appropriate height for standing and walking.

C. *Mobility assistance system*

The mobility assistance system allows the robot to move together with the user, providing support in time across various training environments. It features a mobile base equipped with two actuated rear wheels and two passive front omniwheels. The use of omniwheels enhances the robot's ability to transparently track the user's movements, accommodating both rotational and lateral motion of the pelvis during forward walking. Each omniwheel is independently connected to the base via a separate connecting rod, which can rotate around its joint on the base to adjust the wheel's opening width. This adjustable design offers greater flexibility for the user, enabling a more comfortable walking space during training while also allowing easy passage through doorways when transitioning between different environments and facilities.

D. *Control architecture*

1) User following control: DRBA follows the user by tracking the state of the user's pelvis, which is connected to the robotic arm interface, as shown in Fig 2 . The pelvis state (x, y, α), defined by the center position and orientation of the interface, is determined using measurements from the robotic arm's encoders:

$$x = \frac{l_1 \cos(\pi - \theta_1) + l_2 \cos(\pi - \theta_1 - \theta_2)}{2} + \frac{l_1 \cos\theta_3 + l_2 \cos(\theta_3 + \theta_4 - \pi)}{2} \tag{1}$$

$$y = \frac{l_1 \sin(\pi - \theta_1) + l_2 \sin(\pi - \theta_1 - \theta_2)}{2} + \frac{l_1 \sin\theta_3 + l_2 \sin(\theta_3 + \theta_4 - \pi)}{2} \tag{2}$$

$$\begin{aligned} \alpha = \operatorname{atan2}(&l_1(\sin(\pi - \theta_1) - \sin\theta_3) + \\ &l_2(\sin(\pi - \theta_1 - \theta_2) - \sin(\theta_3 + \theta_4 - \pi), \\ &l_1(\cos(\pi - \theta_1) - \cos\theta_3) + \\ &l_2(\cos(\pi - \theta_1 - \theta_2) - \cos(\theta_3 + \theta_4 - \pi)) \end{aligned} \tag{3}$$

where $\theta_1, \theta_2, \theta_3$ and θ_4 are the joint angles measured by the encoders, and l_1 and l_2 represent the lengths of the proximal and distal links of the robotic arm on one side respectively.

The user-following control ensures the robot maintains a safe distance from the user while aligning its movement with the user's direction. To prevent collisions, the robot stops if the distance falls below a set safety threshold. If the user moves forward and exceeds this threshold, the robot follows to catch up. Additionally, it continuously adjusts its orientation to stay aligned with the user's heading.

Due to the non-holonomic nature of the differential drive, the robot moves along a turning radius R and a turning angle θ, as illustrated in Fig 2. When the user turns with DRBA, they may either rotate their pelvis or take side steps. These two turning preferences result in two possible turning angles for the robot respectively: α and 2β, where $\beta = \operatorname{atan2}(y, x)$. The robot's actual turning angle θ is a weighted combination of these two angles:

$$\theta = k_1\alpha + 2k_2\beta \tag{4}$$

Here, k_1 and k_2 are tunable parameters that can be adjusted based on the user's turning preference, with the constraint $k_1 + k_2 = 1$.

The turning radius is calculated as:

$$R = \frac{|\vec{OP}|\sin(\frac{\pi}{2} - \theta + \beta)}{\sin\theta} \tag{5}$$

where $|\vec{OP}| = \sqrt{x^2 + y^2}$ represents the distance from the robot to the user.

The desired linear velocity v of the robot is proportional to the arc length determined by the turning radius and angle:

$$v = k_p R\theta \tag{6}$$

where k_p is the tunable proportional gain.

Accordingly, the left and right wheel speeds, ω_l and ω_r, can be computed respectively as:

$$\omega_l = \frac{k_p\theta(R - \frac{L}{2})}{r}, \quad \omega_r = \frac{k_p\theta(R + \frac{L}{2})}{r} \tag{7}$$

where L is the distance between the two wheels and r is the wheel radius.

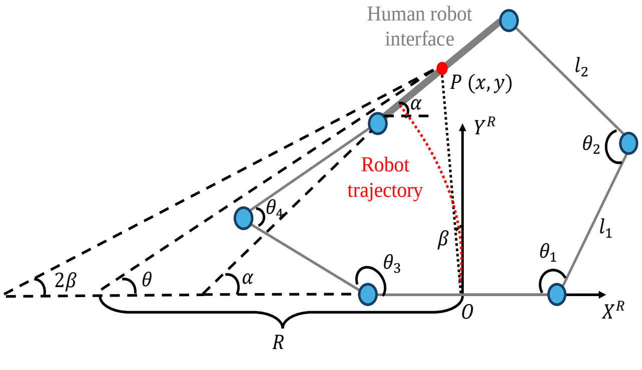

Fig. 2. Schematic diagram of the user following control algorithm. X^R-O-Y^R denotes the robot frame. Joint angles $\theta_1, \theta_2, \theta_3$ and θ_4 are measured by encoders. The lengths of the proximal and distal links of the robotic arm on one side are denoted by l_1 and l_2. Point P represents the center of the human-robot interface, with coordinates (x, y) defined in the robot's frame and the orientation of the interface is given by α. β is the angle between $\vec{OP}$ and Y^R axis. θ and R represent the actual turning angle and turning radius of the robot.

2) *Instability detection and fall intervention:* The instability detection and fall intervention algorithm aims to identify potential fall events during gait training and determine the appropriate moment to intervene, ensuring the user's safety. DRBA continuously monitors the user's stability by tracking body weight transferred to the robotic arm via two force sensors.

The core principle for fall detection is to identify significant changes in supporting forces within a short time frame. Using a sliding window of size T, the force measurements from the left and right sensors are denoted as F_l and F_r respectively. The changes in forces during this window are calculated as: $\Delta F_l = F_l(T) - F_l(0)$ and $\Delta F_r = F_r(T) - F_r(0)$. The state of the user S_{user} is determined based on these force changes accordingly:

$$S_{user} = \begin{cases} \text{Lateral fall} & \text{if } \Delta F_l > \epsilon_1 \text{ or } \Delta F_r > \epsilon_1 \\ \text{Downward fall} & \text{if } \Delta F_l + \Delta F_r > \epsilon_2 \\ \text{Stable walking} & \text{otherwise} \end{cases} \tag{8}$$

Here, ϵ_1 and ϵ_2 represent adjustable thresholds for detecting lateral falls and downward falls respectively. These parameters can be fine-tuned during gait training to align with the user's physical abilities and rehabilitation intensity of challenge-based tasks. Once a fall is detected, the robotic arm's brakes are immediately engaged to lock the robotic arm to provide balance support and prevent the fall. After the user regains stability, they can unlock the robotic arm using a designated button to resume gait training.

III. EXPERIMENT

The evaluation of DRBA was conducted from two perspectives: (i) transparency and (ii) effectiveness. The transparency evaluation assesses the extent to which DRBA interferes with users by examining whether DRBA alters the natural gait of healthy individuals who do not require assistance. The effectiveness evaluation examines whether DRBA can improve users' gait and enhance rehabilitation training. This evaluation involves nine elderly patients with balance impairments as seen in Table I to determine if DRBA supports effective gait training. All experiments utilize a markerless motion capture system [20] with a sampling rate of 30 Hz for gait parameter analysis as shown in Fig 3. The study protocol was approved under JK-2023A-05 by the Institutional Review Board of Guangdong Jianxiang Hospital Group.

A. *Transparency evaluation*

A healthy male participant with no known gait or balance impairments was recruited. The subject completed two experimental conditions designed to compare natural walking patterns with those observed when using DRBA: (1) free walking - the participant was instructed to walk a straight path independently at his preferred walking speed. To capture a comprehensive profile of natural gait variability, the subject completed five walking trials, deliberately varying his speed from slow to his maximum comfortable pace across the trials. (2) walking with DRBA - the participant completed

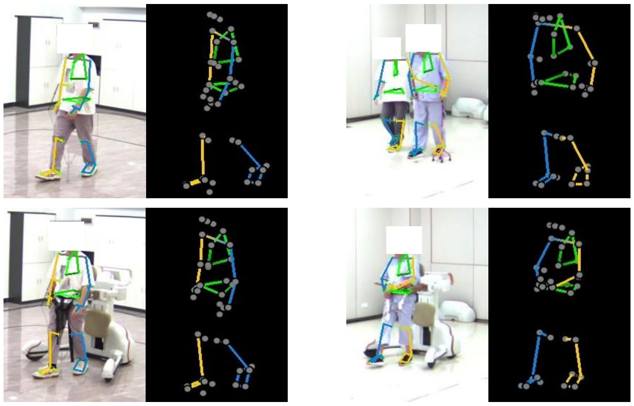
(a) (b)

Fig. 3. Experiment setup. (a) Transparency evaluation. Upper: healthy subject free walking. Lower: healthy subject walking with DRBA. (b) Effectiveness evaluation. Upper: patient assisted by a therapist. Lower: patient assisted by DRBA.

an additional five walking trials along the same path while using DRBA. Prior to data collection, the participant was provided with sufficient time to be familiarized with DRBA, ensuring an adequate adaptation period to minimize learning effects.

B. *Effectiveness evaluation*

Nine elderly patients with various medical conditions and balance impairments were recruited for the trial. Their balance impairments were evaluated using the Berg Balance Scale (BBS) [21]. Participants were categorized into three functional groups according to their BBS scores: (i) Severe balance impairment (0-20) - 3 subjects. (ii) Moderate balance Impairment (21-40) - 5 subjects. (iii) Mild to no balance impairment (41-50) - 1 subject. Table I summarizes the demographic and clinical characteristics of the recruited subjects.

TABLE I

DEMOGRAPHIC AND CLINICAL CHARACTERISTICS OF SUBJECTS

Subject	Gender	Age	Disease type	BBS score
1	Female	84	Stroke	26
2	Male	87	Stroke	50
3	Female	69	Stroke	23
4	Female	76	Parkinson & spondylolisthesis	8
5	Male	75	Stroke	40
6	Female	76	Stroke	22
7	Female	79	Stroke	35
8	Female	76	Stroke & Parkinson	9
9	Male	88	Stroke	20

Patients underwent two distinct walking trials designed to evaluate DRBA's impact on their gait performances:

(1) 10-meter walk assisted by therapist. In this trial, subjects were instructed to walk a straight 10-meter path as quickly as possible while utilizing a crutch and receiving manual assistance from a therapist when necessary. This condition served as a baseline for traditional, therapist-assisted walking without robotic intervention.

(2) 10-meter walk assisted by DRBA. The subjects completed the 10-meter walk under DRBA-assisted condition without direct assistance from the therapist. Prior to data collection, participants received guided training sessions to familiarize themselves with DRBA's operation. Therapists facilitated the initial walking sessions, progressively reducing their level of assistance as participants adapted to DRBA. The level of robotic assistance was tuned individually to suit each participant's specific needs.

IV. RESULTS AND DISCUSSION

A. *Transparency*

The lower limb joint kinematics of the healthy subject in the sagittal plane were analyzed to evaluate the interference introduced by DRBA. Kinematic data from four complete gait cycles were selected from each trial under both free walking and DRBA-assisted walking conditions for comparison. The resulting joint trajectories are presented in Fig 4.

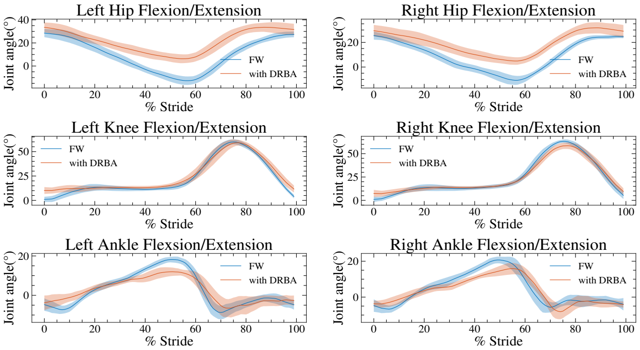

Fig. 4. Comparison of Lower Limb Joint Kinematics in free Walking and DRBA-Assisted walking. The graphs depict sagittal plane joint angles for the hip, knee, and ankle over a complete gait cycle. 'FW' represents free walking without the robot, while 'with DRBA' indicates walking with robotic assistance. Shaded areas denote the standard deviation of joint angles across multiple gait cycles.

The figure reveals a reduced range of motion (ROM) in the hip joint during DRBA-assisted walking, particularly characterized by a noticeable decrease in hip extension. Although DRBA was designed to provide sufficient space in the lower limb region to accommodate the user's natural range of motion, healthy subjects tend to consciously reduce their step length and hip extension when adapting to DRBA. This precautionary adjustment arises from an attempt to avoid potential collisions with the robot, especially since the user cannot directly perceive the device positioned behind them. Additionally, a mismatch in DRBA tuning may also contribute to the reduced hip extension. DRBA is primarily tuned for patients with a typical maximum walking speed below 0.8 m/s. This is supported by gait speed analysis, which showed the subject slowing down to 0.822 ± 0.074 m/s to adapt to the robot.

For the knee joint, no significant differences in the range of motion were observed between free walking and DRBA-assisted walking, indicating minimal interference from DRBA at this joint. However, in the ankle joint, a slight reduction in ankle dorsiflexion was noted just before toe-off (60% of the gait cycle). This minor alteration appears to be a secondary effect of the reduced hip extension and shorter step length.

Furthermore, the subject maintains good gait symmetry while walking with DRBA. The left and right step lengths during DRBA-assisted walking are 0.440 ± 0.034 m and 0.454 ± 0.032 m, respectively, while the left and right step times are 0.541 ± 0.037 s and 0.546 ± 0.035 s. Additionally, no significant inter-cycle gait variations were observed when comparing DRBA-assisted walking to free walking, as demonstrated by the shaded regions of the joint trajectories in Figure 4. This suggests that the user maintains consistent gait stability while walking with DRBA, indicating that the robot introduces minimal interference and does not distort natural gait dynamics.

In summary, DRBA effectively preserves the natural range of motion in the user's lower limb joints. We believe that the minor deviations observed are primarily a result of conscious adaptive strategies employed by the user, rather than direct mechanical interference from the robot, as well as the user deliberately reducing speed to match the patient tune of DBRA. Furthermore, DRBA introduces minimal external influence and does not distort the user's natural gait dynamics. These findings collectively indicate a high level of transparency, demonstrating that DRBA operates with minimal disruption to the user's typical movement patterns.

B. *Effectiveness*

The step length and walking speed of the nine elderly patients were analyzed to assess whether DRBA-assisted training enhances gait performance compared to traditional therapist-assisted training. The improvement in step length and walking speed across all subjects is presented in Figure 5. During the trial, Subject 8 was unable to walk independently with DRBA, despite substantial facilitation from the therapist. This condition was attributed to the subject's lack of gait training for the past two years, leading to severely deteriorated balance perception and a high dependency on therapist assistance when attempting to walk with DRBA. As a result, subject 8 was excluded from the gait improvement analysis as the influence of therapist assistance could not be eliminated.

The results indicate that all subjects exhibited improvements in step length and walking speed when assisted by DRBA compared to therapist-assisted training. However, our patient trials did not provide statistically significant evidence that the BBS score predicts the total gait improvement achieved with DRBA. Pearson correlation analysis yielded a coefficient of -0.25 (p = 0.553), suggesting a weak negative relationship that was not statistically significant. To further explore which users benefit most from DRBA-assisted gait training, we conducted a case-by-case analysis incorporating Manual Muscle Testing (MMT) scores [22]. Notably, Subject 4 had a low BBS score due to multiple medical conditions but retained moderate lower limb muscle strength (MMT: 3–4). Despite relying heavily on DRBA's straps and handles for body weight support, this reduction in balance demands allowed her lower limb muscles to function more effectively, contributing to gait improvement. A similar pattern was observed in the moderate balance impairment group, particularly with Subject 3, who had MMT scores above 4 in all lower limb muscles. Although her BBS score was near the threshold for severe balance impairment, her strong and well-distributed muscle function resulted in the greatest gait improvements among all subjects. In contrast, Subject 6, who had lower limb muscle strength ranging from MMT 2- to 4, struggled due to severe ankle dorsiflexion limitations. Even with DRBA's body weight support, these functional deficits restricted her walking ability.

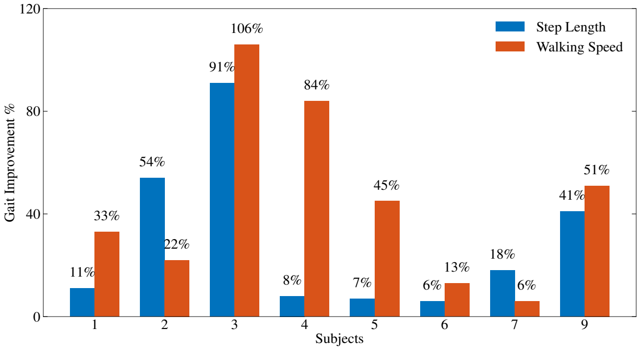

Fig. 5. Gait Improvement in DRBA-Assisted vs. therapist-assisted training. The graph presents the percentage improvement in step length (blue bars) and walking speed (red bars) for all subjects during DRBA-assisted training compared to traditional therapist-assisted training.

Observations from the patient trial also indicated that most subjects adapted to DRBA within a few sessions under therapist facilitation. Once adapted, they could walk independently with DRBA, requiring minimal safety concerns and only occasional therapist assistance. The improvements in gait and the expanded range of activity options increased their engagement and motivation in rehabilitation sessions. Additionally, DRBA reduced the therapist's supervision burden, enabling a single therapist to oversee multiple patients simultaneously in a single rehabilitation session.

Overall, our patient trials have shown that DRBA-assisted training enhances gait rehabilitation by increasing training dosage through effective and adaptive body weight support. Preliminary findings indicate that some patients achieved 70–80% of their maximum heart rate during DRBA training, highlighting its potential for high-intensity rehabilitation. Notably, patients with severe balance impairments but good lower limb muscle strength benefit the most, as DRBA reduces balance-related effort, allowing them to focus on walking mechanics rather than stability maintenance, a key limitation of traditional therapist-assisted gait training.

Moreover, DRBA enables users to engage in tasks beyond their unaided physical capacity, promoting functional and challenge-based rehabilitation through its flexible and transparent pHRI design. The perceived safety and effective assistance provided by DRBA helped users build confidence, alleviating mental barriers and encouraging more frequent and prolonged participation in rehabilitation sessions.

V. CONCLUSIONS AND FUTURE WORK

This paper introduces the Dynamic Robotic Balance Assistant (DRBA), a novel gait rehabilitation robot designed for diverse training environments. DRBA features a compliant 3-DoF robotic arm for assist-as-needed pelvic support, ensuring minimal interference during stability while providing timely assistance during instability. Its flexible design enables functional and challenge-based rehabilitation, broadening training possibilities. A compact sit-to-stand assistance system facilitates safe and seamless transitions, while the mobility assistance system allows DRBA to move synchronously with the user. Additionally, A user-following algorithm and sensitive fall detection system further ensure minimal disruption to natural movement while delivering responsive balance support when needed.

Experimental results demonstrated that DRBA maintains high transparency, with minimal impact on users' gait patterns and stability. Pilot studies with nine patients (BBS: 8–50) across various medical conditions confirmed its effectiveness, showing increased step length and walking speed compared to therapist-assisted training. Preliminary data indicated that some patients reached 70–80% of their maximum heart rate during DRBA training, highlighting its potential for high-intensity rehabilitation. DRBA also enabled users to perform tasks beyond their unaided physical capacity, expanding rehabilitation options and training diversity. The perceived safety and increased activity choices fostered greater engagement in rehabilitation sessions. These findings suggest that DRBA can enhance rehabilitation outcomes by facilitating higher training intensity, promoting frequent and prolonged sessions, and enabling task-oriented exercises tailored to individual needs, in alignment with the FITT principle.

Future research will focus on conducting long-term clinical trials to further evaluate DRBA's rehabilitation impact and explore its effectiveness across different patient populations. We also aim to expand task-specific training protocols to optimize its therapeutic potential. Additionally, usability studies will be conducted to assess user experience, safety, and ease of integration into rehabilitation settings, ensuring that DRBA meets the practical needs of both patients and therapists.

ACKNOWLEDGMENT

This research is supported by the Singapore National Robotics Programme (NRP) BAU grant - Mobile Robotic Balance Assistant (Award No: M23NBK0045). The authors extend their gratitude to Shiwei Wu, Zhi Wang, Yushu Wang, Chen Tian, and Junhua Liang from the Guangdong Zhongxin Intelligent Rehabilitation Research Institute for their valuable technical assistance and support in experiment setup.

REFERENCES

[1] L. Nyberg and Y. Gustafson, "Patient falls in stroke rehabilitation: a challenge to rehabilitation strategies," *Stroke*, vol. 26, no. 5, pp. 838–842, 1995.

[2] M. A. KEENAN, J. PERRY, and C. JORDAN, "Factors affecting balance and ambulation following stroke." *Clinical Orthopaedics and Related Research (1976-2007)*, vol. 182, pp. 165–171, 1984.

[3] J. Desrosiers, L. Noreau, A. Rochette, G. Bravo, and C. Boutin, "Predictors of handicap situations following post-stroke rehabilitation," *Disability and rehabilitation*, vol. 24, no. 15, pp. 774–785, 2002.

[4] C.-L. Lin, S.-F. Hsieh, M.-H. Hsiao, and J.-H. Huang, "Predicting long-term care institution utilization among post-rehabilitation stroke patients in taiwan: a medical centre-based study," *Disability and Rehabilitation*, vol. 23, no. 16, pp. 722–730, 2001.

[5] A. S. Pollock, B. R. Durward, P. J. Rowe, and J. P. Paul, "What is balance?" *Clinical rehabilitation*, vol. 14, no. 4, pp. 402–406, 2000.

[6] J. Mehrholz, S. Thomas, and B. Elsner, "Treadmill training and body weight support for walking after stroke," *Cochrane database of systematic reviews*, no. 8, 2017.

[7] S. Jezernik, G. Colombo, T. Keller, H. Frueh, and M. Morari, "Robotic orthosis lokomat: A rehabilitation and research tool," *Neuromodulation: Technology at the neural interface*, vol. 6, no. 2, pp. 108–115, 2003.

[8] D. Marks, R. Schweinfurther, A. Dewor, T. Huster, L. P. Paredes, D. Zutter, and J. C. Möller, "The andago for overground gait training in patients with gait disorders after stroke-results from a usability study," *Physiother Res Rep*, vol. 2, no. 2, pp. 1–8, 2019.

[9] B. Leme, M. Hirokawa, H. Kadone, and K. Suzuki, "A socially assistive mobile platform for weight-support in gait training," *International Journal of Social Robotics*, vol. 13, pp. 459–468, 2021.

[10] H. Vallery, P. Lutz, J. von Zitzewitz, G. Rauter, M. Fritschi, C. Everarts, R. Ronsse, A. Curt, and M. Bolliger, "Multidirectional transparent support for overground gait training," in *2013 IEEE 13th International Conference on Rehabilitation Robotics (ICORR)*. IEEE, 2013, pp. 1–7.

[11] M. Frey, G. Colombo, M. Vaglio, R. Bucher, M. Jorg, and R. Riener, "A novel mechatronic body weight support system," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 14, no. 3, pp. 311–321, 2006.

[12] J. Patton, D. A. Brown, M. Peshkin, J. J. Santos-Munné, A. Makhlin, E. Lewis, E. J. Colgate, and D. Schwandt, "Kineassist: design and development of a robotic overground gait and balance therapy device," *Topics in stroke rehabilitation*, vol. 15, no. 2, pp. 131–139, 2008.

[13] A. A. Morbi, "Design, control, and implementation of a robotic gait rehabilitation system for overground gait training," Ph.D. dissertation, Carleton University, 2014.

[14] K.-R. Mun, H. Yu, C. Zhu, and M. S. Cruz, "Design of a novel robotic over-ground walking device for gait rehabilitation," in *2014 IEEE 13th international workshop on advanced motion control (AMC)*. IEEE, 2014, pp. 458–463.

[15] G. Aguirre-Ollinger and H. Yu, "Omnidirectional platforms for gait training: Admittance-shaping control for enhanced mobility," *Journal of Intelligent & Robotic Systems*, vol. 101, no. 3, p. 52, 2021.

[16] K.-R. Mun, S. B. Lim, Z. Guo, and H. Yu, "Biomechanical effects of body weight support with a novel robotic walker for over-ground gait rehabilitation," *Medical & biological engineering & computing*, vol. 55, pp. 315–326, 2017.

[17] L. Li, M. J. Foo, J. Chen, K. Y. Tan, J. Cai, R. Swaminathan, K. S. G. Chua, S. K. Wee, C. W. K. Kuah, H. Zhuo, *et al.*, "Mobile robotic balance assistant (mrba): a gait assistive and fall intervention robot for daily living," *Journal of NeuroEngineering and Rehabilitation*, vol. 20, no. 1, p. 29, 2023.

[18] Y. Wang, M. Yuan, L. Li, K. S. G. Chua, W. T. Ang, *et al.*, "Graceful user following for mobile balance assistive robot in daily activities assistance," *IFAC-PapersOnLine*, vol. 56, no. 2, pp. 1139–1144, 2023.

[19] A. C. of Sports Medicine *et al.*, *ACSM's guidelines for exercise testing and prescription*. Lippincott williams & wilkins, 2013.

[20] P. Jatesiktat, G. M. Lim, W. S. Lim, and W. T. Ang, "Anatomical-marker-driven 3d markerless human motion capture," *IEEE Journal of Biomedical and Health Informatics*, 2024.

[21] K. Berg, S. Wood-Dauphine, J. Williams, and D. Gayton, "Measuring balance in the elderly: preliminary development of an instrument," *Physiotherapy Canada*, vol. 41, no. 6, pp. 304–311, 1989.

[22] S. C. Cuthbert and G. J. Goodheart Jr, "On the reliability and validity of manual muscle testing: a literature review," *Chiropractic & osteopathy*, vol. 15, no. 1, p. 4, 2007.