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Gyroscopic wearable improves balance performance in people with degenerative ataxia – a sham-controlled robotics study

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Degenerative ataxias cause progressive cerebellar dysfunction, leading to gait and balance impairments. Patients are often dissatisfied with current assistive devices, such as walkers. We developed the GyroPack, a gyroscopically-actuated, balance-assisting backpack, and evaluated its effects in 14 individuals with degenerative ataxia using a single-blinded, sham-controlled study. Participants completed five static and dynamic balance tasks under three conditions: baseline, sham, and assistive. In assistive mode, the GyroPack dampened trunk rotation; in sham mode, this effect was minimal. The GyroPack increased the average standing time when comparing assistive to baseline condition. While walking, it also reduced the variability of the trunk angular velocity and the extrapolated center of mass, both indicators for gait stability. When comparing assistive to sham, there was an overall trend for differences, however, only trunk angular velocity differed significantly. These findings mark an important step toward a portable robotic wearable for individuals with cerebellar and other neurological conditions.

Degenerative ataxias are a group of progressive diseases that affect the cerebellum[1]. Gait and balance impairments are among the most disabling symptoms in people with degenerative ataxias[2], as these result in reduced mobility, falls and fall-related injuries, reduced participation in society, and a markedly reduced quality of life[3–5]. The gait patterns observed in people with ataxia are often unstable with high movement variability[6–8]. Standing also becomes unstable in moderate to advanced stages of ataxia[7,8]. Wheeled walkers, often with extra weight, are commonly used in moderate to advanced stages, to compensate for reduced balance capacity and to prevent falling[5,9]. However, their bulky designs make them less convenient to use in narrow spaces, and the heavy weight makes them difficult to lift over obstacles. Patients often ask for alternatives, which are currently lacking.

We have developed a technological alternative: a gyroscopically-actuated balance-assisting backpack (GyroPack). The GyroPack consists of control moment gyroscopes, an actuation principle typically deployed in space applications, which relies on manipulating the angular momentum of fast-spinning flywheels[10]. This device can be controlled such that it exerts a moment counteracting the angular velocity of the trunk and thereby dampening trunk motion. This moment is considered to be a "free" moment because it does not require a connection or anchor to an inertial reference frame, such as the floor or wall, to generate a reaction moment. This advantage implies that the wearer can move freely in 3D space[11,12].

An earlier version of the device, which was much heavier and equipped with only a single control moment gyroscope, was tested in a case series with post-stroke individuals[13]. We showed that static balance performance, when standing on a beam, improved with the use of the gyroscopic backpack[12]. However, we could not effectively blind participants to the different conditions, and the device required external weight support, limiting its implementation in daily life. Therefore, we developed a lighter iteration: this device weighs only six kilograms and contains two control moment gyroscopes[14]. We found that this novel device reduced trunk motion variability in healthy subjects and improved static and dynamic balance[11]. With two control moment gyroscopes, it is possible to create a sham condition, where the flywheels have opposing angular momentum vectors. While placebo/sham conditions are ubiquitous in other research paradigms, they are difficult to emulate for assistive devices given the current designs

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Table 1 | Demographics and clinical characteristics of participants

Number of participants		14
Age (years)		57 (±7)
Sex (female)		2
Weight (kg)		89 (±15)
Height (cm)		185 (±11)
Diagnosis (number of participants)		
Clinical	ILOCA	1
	ADCA, no genotype	2
	CANVAS, no genotype	1
Molecular	SCA-1	4
	SCA-3	3
	SCA-6	1
	SCA-14	1
	SCA-28	1
Disease duration (years)		16 [4–52]
Activities-specific Balance Confidence Scale (%)		55 [42–96]
Scale for the Assessment and Rating of Ataxia (points)		12 [7–20]
Mini-Balance Evaluation Systems Test (points)		14 [4–25]

Values displayed are means (±SD) or median [range]. SCA Spinocerebellar ataxia, ILOCA Idiopathic late-onset cerebellar ataxia, ADCA Autosomal dominant cerebellar ataxia, CANVAS Cerebellar ataxia, neuropathy and vestibular areflexia syndrome. Activities-specific Balance Confidence Scale (ABC) ranges between 0–100%; higher scores indicate higher confidence. Scale for the Assessment and Rating of Ataxia (SARA) ranges from 0–40; higher scores indicate higher disease severity. Mini Balance Evaluation Systems Test (Mini-BESTest) ranges from 0–28 points; higher scores indicate better balance capacity.

and methods of actuation. Nevertheless, considering a possible placebo-effect is important for a fair evaluation of a device's efficacy.

In the current study, we aimed to provide first proof-of-principle evidence of the efficacy of the GyroPack to improve balance performance in persons with moderate to advanced degenerative ataxia. We used a single-blinded, sham-controlled cross-sectional study design.

Results

We recruited fourteen individuals with degenerative ataxia (see Table 1 for patient characteristics) to test the efficacy of the GyroPack (see Figs. 1A and 1B). On one single day, participants performed five tasks: T1 – standing, eyes open, with feet together, T2 – walking for two minutes, T3 – 360° turns-in-place, T4 – tandem stance, and T5 –backward and forward treadmill perturbations. T1 and T4 were both capped at 30 seconds. All tasks were performed under three conditions: without the device (i.e., *baseline)*, with the device in *sham* mode, and with the device in *assistive* mode. During *assistive* mode, the device was configured as a rotational damper, see Fig. 1-C. During *sham* mode, the damping effect was reduced to a minimum, while motor noises remained similar to *assistive* mode, see Fig. 1-D. An example video recording several tasks is available in supplementary materials Movie S1. The frequently elicited Borg Rating of Perceived Exertion scale remained stable around ten, indicating that the perceived exertion was light.

Damping power during assistive and sham mode

We evaluated the functionality of the backpack by investigating its damping power and torque profiles. Trunk angular velocity and torque profiles of the device during sham and assistive modes are shown in Fig. 1C, D. For the assistive mode, Fig. 1C shows an inverse linear relationship between trunk angular velocity and backpack torque, i.e., an increase in rotational velocity coincides with an increase in opposing (negative) rotational torque. For the sham mode, Fig. 1D shows that such a clear relationship is absent, i.e., there

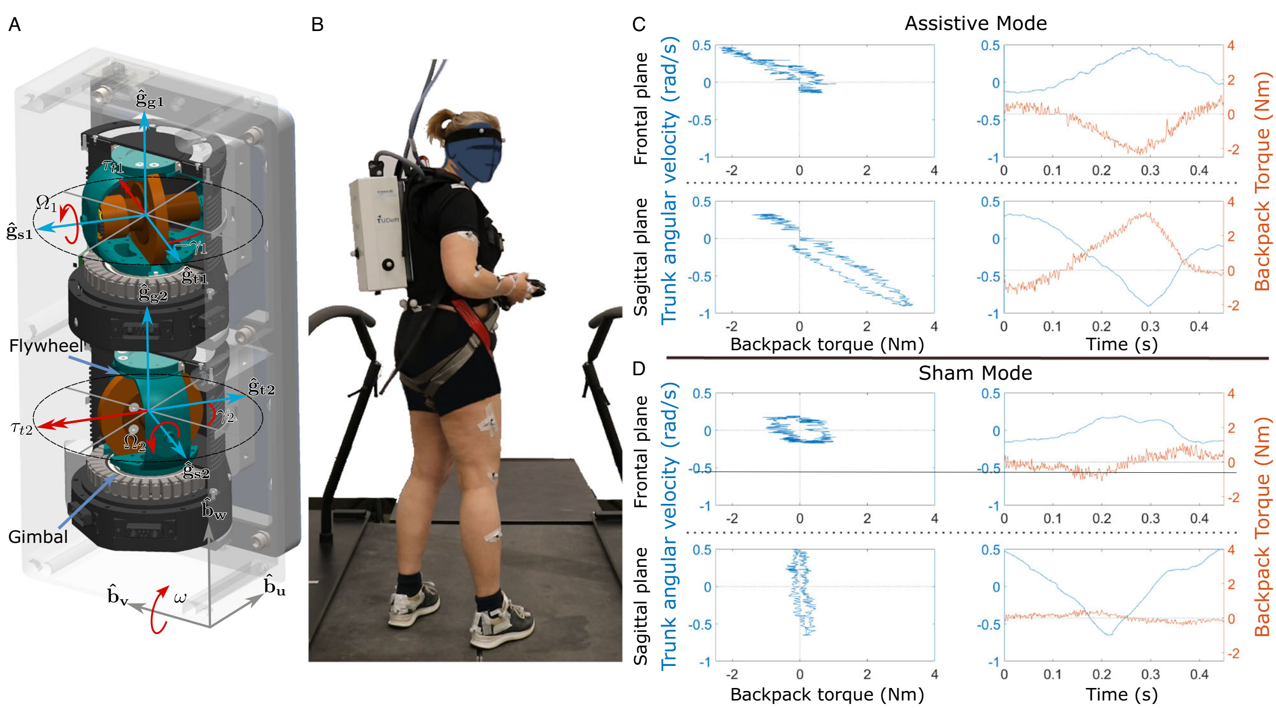

Fig. 1 | Balance assisting robotic wearable with gyroscopic actuators. A A 3D rendering of the robotic wearable, with a cross-section of the two stacked control moment gyroscopes, each containing a flywheel and gimbal. **B** A participant wearing the GyroPack. **C, D** Trunk angular velocity and torque profiles of the device during *assistive* and *sham* mode, both recorded during a walking trial. Only in assistive mode, the control algorithm explicitly enforced a linear (damping) relationship between angular velocity and backpack torque.

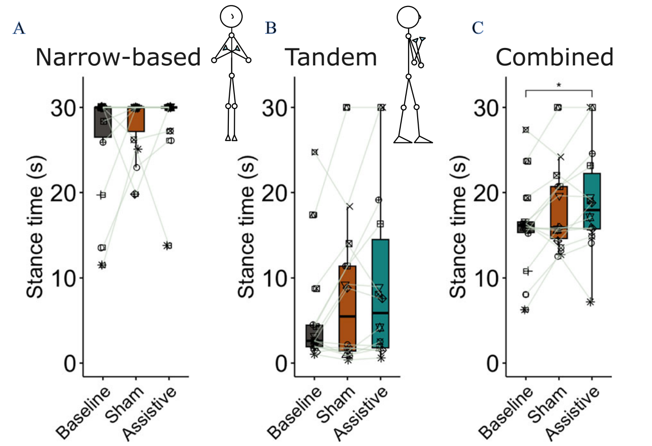

Fig. 2 | Stance times for narrow-based and tandem stance, and their average. Three boxplots showing the stance time of: **A** standing with feet together, eyes open, and arms crossed (T1), **B** tandem stance, eyes open, with arms crossed (T4), and **C** the combined average of T1 and T4. An * above the bars indicates a significance level of $p < 0.05$. The outcomes of each participant are displayed with a unique symbol.

is no one-to-one mapping of velocity to torque. The real-time controller updates the torque setpoint, based on the measured angular velocity, at a rate of 1000 Hz, and the actuator bandwidth is approximately 18 Hz[14], such that the delay between velocity and torque is considered negligible compared to the slow system dynamics of human balance.

During the walking trials, the GyroPack applied an average power of −0.02 (±0.01) W in the *sham* mode and −0.19 (±0.08) W in the *assistive* mode, indicating an eight-fold increase in overall damping power. The highest damping peaks across all trials were −1.55 W for *sham* mode and −3.89 W for *assistive* mode. The maximum peak torque recorded during any of the trials was 4.35 N m (during *assistive* mode).

Static standing balance

For both the narrow-based stance (T1) and the tandem stance (T4) tasks, a non-parametric Friedman test showed no significant differences in stance times between conditions. However, stance times in the narrow-based stance condition displayed a ceiling effect in a subgroup of participants (Fig. 2A), whereas tandem stance outcomes displayed floor effects (Fig. 2B) in others. For each participant, we therefore also calculated the combined average stance time of T1 and T4, for which the Wilcoxon signed-rank test showed that the *assistive* mode (19.0 s ± 6.2) increased stance times compared to *baseline* condition (16.0 s ± 5.5; $p = 0.02$, $r = 0.696$). No significant differences were found between *assistive* mode and *sham* mode (18.1 s ± 5.0; $p = 0.62$, $r = 0.336$), and between *sham* mode and *baseline* condition ($p = 0.20$, $r = 0.495$), nor for the non-parametric Friedman test.

The Root Mean Square (RMS) of the Center of Pressure (CoP) velocity, evaluated in the narrow-based stance task (T1), also differed between conditions (Friedman $\chi^2 = 8.143$, $p = 0.017$, as shown in supplementary Figure S1). Post-hoc Wilcoxon signed-rank tests showed that the device in *assistive* mode (0.036 m/s ± 0.021) reduced the RMS of the CoP velocity compared to *baseline* (0.048 m/s ± 0.031; $p = 0.05$, r = 0.629), whereas no significant differences were found between *assistive* mode and *sham* mode (0.041 m/s ± 0.027; $p = 0.40$, $r = 0.411$), and between *sham* mode and *baseline* condition ($p = 0.80$, $r = 0.310$).

Dynamic walking variability

During the walking task (T2), the trunk angular velocity variability differed between conditions (Friedman $\chi^2 = 11.231$, $p = 0.0036$, Fig. 3A). Post-hoc Wilcoxon signed-rank tests showed that the device in *assistive* mode (4.54 ± 0.59) reduced trunk angular velocity variability compared to *baseline* (5.37 ± 0.89; $p = 0.001$, $r = 0.863$), and compared to *sham* mode (4.92 ± 0.75; $p = 0.032$, $r = 0.688$), whereas no significant differences were found between *sham* mode and *baseline* condition ($p = 0.051$, $r = 0.649$).

Step width variability, as shown in Fig. 3B, showed no significant differences across conditions. Similarly, the RMS of the mediolateral CoM position, as shown in Fig. 3C, also showed no significant differences between conditions. The RMS of the mediolateral extrapolated center of mass (xCoM) differed between conditions (Friedman $\chi^2 = 14$, $p < 0.001$, Fig. 3D). Post-hoc Wilcoxon signed-rank tests showed that the device in *assistive* mode (0.35 ± 0.12) reduced the RMS of the mediolateral xCoM compared to *baseline* (0.39 ± 0.13; $p = 0.004$, $r = 0.824$), and between *sham* mode (0.36 ± 0.13) and a *baseline* condition ($p < 0.001$, $r = 0.882$), whereas no significant differences were found between *assistive* mode and *sham* mode ($p = 1$, $r = 0.223$).

Turning, recovery after perturbations, and perceived assistance

For the turns-in-place task (T3), a non-parametric Friedman test showed no significant differences in average turning velocities across conditions (Friedman $\chi^2 = 3.857$, $p = 0.15$). During the perturbation recovery task (T5), we measured the number of successful feet-in-place recoveries, but the differences were not significant (14.3% (baseline), 18.9% (sham), and 20.5% (assistive), (Friedman $\chi^2 = 1.52$, $p = 0.47$)). The participants' subjective perception of the damping effect was measured using a 5-point Likert scale across all five tasks, which yielded no differences between the sham (3.7 ± 0.5) and assistive (3.8 ± 0.5) mode.

Discussion

We evaluated the effects of a novel gyroscopic balance assisting device (GyroPack) on balance and gait performance in persons with moderate to advanced degenerative ataxia, using a single-blinded, sham-controlled, cross-sectional study. The GyroPack was able to increase the average standing time when comparing *assistive* to *baseline* condition. While walking, it also reduced the variability of the trunk angular velocity and the xCoM, which are frequently used indicators for gait stability[15–17]. Trunk angular velocity variability also showed a significant reduction when comparing *assistive* to *sham*. Performance on perturbation recovery and turns-in-place tasks did not show significant differences across conditions. When observing the means of all outcome measures, in general, a trend is visible for improvement from *baseline* to both *sham* and *assistive*, with further improvements from *sham* to *assistive*.

One of the strengths of the current study is the addition of a *sham* condition, where sensory input, i.e., noise, weight, vibrations, received by the participants was kept as similar as possible compared to the *assistive* mode. This is often omitted during studies on rehabilitative technology, as demonstrated by recent systematic reviews[18,19]. At first sight, it may be surprising that participants performed better during the *sham* condition compared to the *baseline* condition during the walking task, as evidenced by reduced variability of the mediolateral xCoM. The reasons might be threefold. First, most likely there is an effect of the added weight to the torso. Indeed, studies showed that adding weight to the torso (i.e., "torso weighting") can help reduce sway velocity, improve standing time and directional imbalance in people with ataxia[20,21]. Second, as shown by the average power calculations during the walking trials, the damping power of the *sham* condition was not absolute zero, but approximately one eighth of the damping power during the *assistive* condition. This, in combination with the relatively high damping peaks, might indicate that the backpack contributed more assistance than desired to the gains in balance performance. Third, it might be a true placebo effect. The perceived assistance score of the *sham* condition also indicates that the participants perceived that the backpack was providing as much assistance as during the *assistive* condition. Placebo responses have previously been demonstrated on various objective ataxia scales, including significant improvements of the SARA score subitems gait and stance[22,23].

When comparing *assistive* to *sham* mode, the outcome measures showed trends of improvement, however, only trunk angular velocity variability was significantly reduced. As weight, noise, and user perception

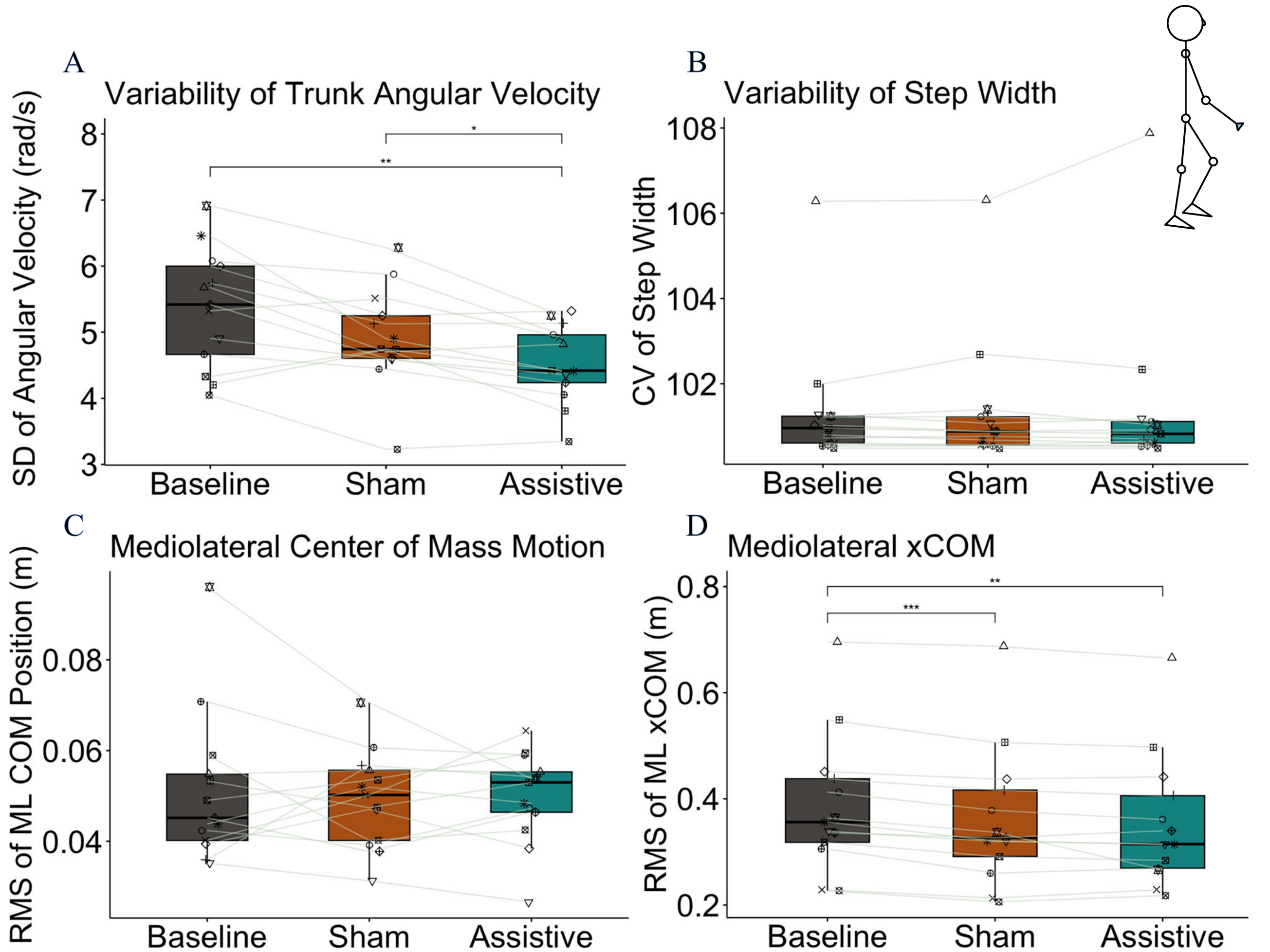

Fig. 3 | Trunk angular velocity and xCOM variability significantly reduced while wearing the GyroPack during walking. Boxplots showing (**A**) the standard deviation of trunk angular velocity, (**B**) the coefficient of variance of step width, and the Root Mean Squares of the (**C**) mediolateral Center of Mass (CoM) position and the (**D**) extrapolated CoM, across one minute of walking, for *baseline*, *sham*, and *assistive* conditions. An * above the bars indicates a significance level of $p < 0.05$, ** indicates a significance level of $p < 0.01$, and *** indicates a significance level of $p < 0.001$. The outcomes of each participant are displayed with a unique symbol.

stayed the same, this improvement is most likely due to the eight-fold increase in damping power. These gains were absent during the perturbation and turning tasks. From a technical standpoint, this is likely due to the nature of the standing and walking tasks, where the trunk repeatedly moves from left-to-right or front-to-back, in combination with the actuation principle that is prone to singularities and thereby saturation of the moment vector. During cyclic tasks, the desired torque vector is repeatedly flipped 180 degrees, re-enabling the saturated gyroscopes to generate a damping moment. The perturbation and turns-in-place tasks are unidirectional movements, so the gyroscopes might be saturated, i.e., depleted, before they are able to provide meaningful assistance. Additionally, the device only provides rotational damping around the frontal and sagittal axis of the trunk, thus it provides little protection against the translation of the COM during the perturbation task, nor against the rotation around the vertical axis during the turns-in-place task.

When observing which type of outcome measure showed improvements, the highest gains were displayed for velocity-based metrics, e.g., trunk angular velocity, COP velocity, and xCOM, which, from a technical standpoint, makes sense as the device was configured as a damper, which is velocity-based. The absence of improvements of the ML COM excursions might be explainable as the device only dampens rotation, not translation. Clinically, the absence of gains in, for instance, step-width variability is explainable as the device does not resolve lower limb ataxia[24]. One advantage of the current – damping – controller is that it does not require any knowledge of the user's desired orientation with respect to the reference. Although this should be demonstrated in further studies, this means that the device could also be used during more complex tasks, such as sit-to-stand, running, or crouching maneuvers.

The improvements in balance and gait performance were relatively small and currently do not outweigh the device's disadvantages for application in daily life. The main limitations are its weight, noise, and low damping power. The issue of weight is an inherent problem of this type of technology, as the available angular momentum, and thus output torque, is proportional to flywheel mass. The noise of the device during the trials was about 86 dB, whereas a single gyroscope in our laboratory setup only produces about 75 dB. This difference is due to the resonance interaction between the gyroscopes and poor design of the outer protective casing and suspension. Because of this, we also had to reduce the flywheel speed, and thus available output torque, of one of the gyroscopes by about 10%. In previous explorations, with an overpowered device, the exerted peak moments ranged on the order of 10-15 N m[12], whereas in the current study the maximum was 4.3 N m. We estimate that with redesign the device's weight can be reduced to five kilograms, and the noise to 78 dB. Further gains in torques might be possible but are hard to estimate.

In its current form, the GyroPack is mostly useful for research and clinical training, due to its flexibility in configurations. The device can be reconfigured to render different damping strengths, including a negative damper, for applications in training strategies such as "error augmentation,"

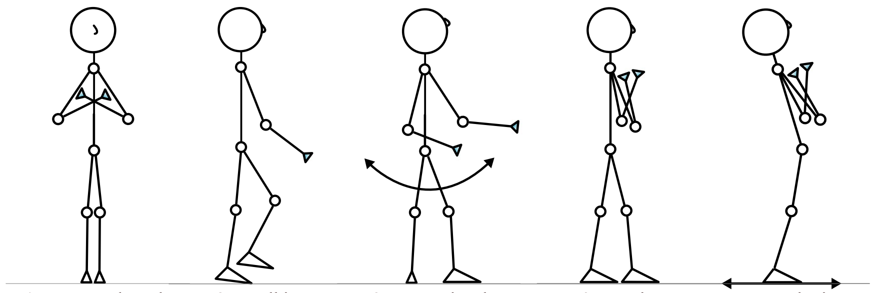

Fig. 4 | Five static and dynamic balance tasks were executed and evaluated. Stick figures displaying the five tasks that participants had to perform during *baseline*, *sham* and *assistive* conditions. Tasks were executed from left to right.

which aims to enhance learning by increasing feedback error[25,26]. It could also be used in rendering moments in telerehabilitation or virtual reality environments[27], or for perturbation training.

Further investigation should investigate task-specific controllers, where the actuators adjust their orientation in anticipation of disturbances in directions that are likely to be more challenging. For example, during the tandem stance task (T4), more support in the ML direction may be needed, while the narrow-based stance task (T1) might benefit more from an omni-directional controller. Overall, this work represents an important step toward the development of a fully portable robotic wearable for assisting balance in people with gait and balance impairments.

Methods

Study design

This is a single-blinded, sham-controlled cross-sectional study, involving a single measurement session. The study was approved by the local medical ethical committee (Medisch Ethische Toetsings Commissie (METC) Oost-Nederland (NL 82737.091.22)) and preregistered online at https://osf.io/7ug5j.

Study participants

Fourteen individuals diagnosed with degenerative ataxia were included after signing informed consent and passing the screening criteria. Informed consent was obtained after the nature and possible consequences of the studies were explained. The inclusion criteria were 1) diagnosis of pure degenerative ataxia, based on molecular diagnosis and/or clinical assessment, 2) age between 18-70 years, 3) difficulties with gait where the participant cannot perform tandem walking (heels to toes) over ten steps and staggering occurs while performing a half turn (Score of 2, 3 or 4 on item 1 'Gait' of the Scale for the Assessment and Rating of Ataxia (SARA)[28]), 4) difficulties while standing that arise when the participant can stand >10 seconds, but only with sway; intermittent support of the wall is allowed (Score of 2, 3 or 4 on item 2 'Stance' of the SARA scale), 5) being able to wear a backpack of six kg for a duration of two hours. Participants were excluded if other neurological or orthopedic conditions impacting balance and gait capacity were present.

Device design

The GyroPack is a wearable robotic backpack, which contains two control moment gyroscopes. Each gyroscope can impart a maximum torque impulse of approximately 20 N m for 0.1 seconds in a fixed direction before reaching singularity, at which point no further torque can be generated[29]. A change in desired torque direction re-enables the gyroscope's ability to impart a torque. During the trials in this study, both control moment gyroscopes were mounted with their gimbal axes aligned with the longitudinal axis of the user, allowing the device to exert a moment about the user's frontal and sagittal axes. The inertial measurement units (IMU) inside each gyroscope were used to obtain the user's trunk angular velocity ($\boldsymbol{\omega}$).

Two controllers were developed for the experiments: *assistive* and *sham*. First, as the *assistive* controller a positive rotational damper was implemented. Here, the gyroscopes impart a moment opposite and proportional to the combined frontal axis ($\hat{\mathbf{u}}_u$, sideways rotation) and sagittal axis ($\hat{\mathbf{u}}_v$, forward rotation) components of $\boldsymbol{\omega}$. A torque opposing the trunk's angular velocity is exerted to dampen the trunk's motion. When approaching singularities, the moment is saturated[10,29].

Second, a *sham* controller was implemented, in which the angular momentum vectors of both gyroscopes were maintained at a relative 180 degrees, thereby canceling out their independent effects. To mimic motor movements and noises without directional preference, the gimbal velocities were coupled only to the magnitude of the $\boldsymbol{\omega}$ vector, with a small gain.

The full description of the device and each controller can be found in Supplementary Materials and Methods.

Testing protocol

The clinimetrics that were collected included: 1- demographics, 2- disease severity via the Scale for the Assessment and Rating of Ataxia (SARA)[28], 3- self-perceived balance capacity via the Activities-specific Balance Confidence Scale (ABC) questionnaire[30], and 4- balance capacity assessed via the Mini Balance Evaluation Systems Test (Mini-BESTest)[31].

Subsequently, the participant's preferred walking speed was obtained by walking on the treadmill at increasing speeds until the participants indicated they could comfortably maintain that speed for several minutes. Then, the speed was fine-tuned by lowering and increasing the speed until the final preferred walking speed was found.

Participants performed five tasks, as visualized in Fig. 4, under three conditions: a *baseline* (B) condition without the GyroPack, and *sham* (P) and *assistive* (A) conditions with the GyroPack (with their respective controllers activated). The conditions were block-randomized to limit delays from device donning and doffing, as well as the velocity ramp-up time of the flywheels. The order of conditions was counterbalanced to account for learning and fatigue effects. Half of the participants performed all the tasks in the order B-A-P-P-A-B, whereas the other half performed all tasks in the order B-P-A-A-P-B, both with a mandatory break after the third set of tasks. Participants were offered breaks between conditions and participant fatigue was monitored using the Borg Rating of Perceived Exertion scale by eliciting the participant to indicate their fatigue after every set of tasks[32].

For the first task, participants were instructed to **stand (T1)** as still as possible, with their eyes open, feet together and arms crossed comfortably across the chest. The stance time was measured and capped at 30 seconds, as is the consensus recommendation for this target population[7]. The timer started as soon as the wrists crossed each-other in front of the body and stopped as soon as either: the wrists uncrossed, when a step was made, or when the participant leaned into the safety harness. The primary outcome was the RMS of the low-pass filtered Center of Pressure (CoP) velocity (20 Hz 6th order Butterworth), which was calculated as the RMS of the Euclidean norm of the first derivatives of the ML and AP CoP positions.

Secondly, participants **walked (T2)** on the treadmill at their preferred speed for two minutes, without using the handrail and without relying on the safety tether for support. The last minute was used for analysis to allow 60 seconds for reaching steady-state walking. The ML Center of Mass (CoM) position and extrapolated CoM (xCoM) were estimated based on a combination of ground reaction forces (GRF) and CoP positions[33]. This method was preferred over the simpler approach of estimating the CoM using a centrally placed motion capture marker, because the backpack influences the CoM due to its own weight and complicates marker placement on the participant's torso. Dividing the ML GRF by the participants' weight resulted in the ML CoM acceleration, which was then twice integrated and high-pass filtered (0.2 Hz 2nd order Butterworth) and added to the low-pass filtered ML CoP (0.2 Hz 2nd order Butterworth). The ML xCoM was calculated as follows, where g is the gravitational constant, ℓ is defined as leg length multiplied by 1.2[34], CoM is the position of the CoM, and CoM velocity (vCoM) was calculated during the above-mentioned integration step[34].

$$xCoM = CoM + \frac{vCoM}{\sqrt{g/\ell}} \tag{1}$$

Step width was calculated as the distance, perpendicular to the walking direction, between two subsequent heel strikes. To determine step width variability, the coefficient of variance across all steps was calculated. For analysis of the trunk angular velocity ($\boldsymbol{\omega}$), each stride cycle, defined as the period between two consecutive left heel strikes, was resampled into 100 frames. The average of $\boldsymbol{\omega}$ at each corresponding frame across all stride cycles produced the "average" stride trajectory for each trial. The standard deviation of $\boldsymbol{\omega}$ at each frame, across all strides, was calculated and subsequently averaged across the normalized stride cycle to produce the trunk angular velocity variability[35]. One participant was excluded from the walking task, as they were not able to walk on the treadmill for more than ten seconds without using the handrail or safety-tether.

The third task comprised two consecutive 360° **turns-in-place** (**T3**), clockwise and counterclockwise, with a slight pause in between, without using the handrails or safety tether. Participants were asked to turn by using quick steps (i.e., a pirouette was not allowed). The average rotational speed was used as primary outcome measure. As the turn was self-guided, not all attempts were exact 360° turns, so the average speed was taken over a 330° section of the rotation.

Fourthly, participants performed a **tandem** stance (**T4**), with the heel of one foot touching the toes of the other foot and arms crossed, for as long as possible. Stance time was analyzed as primary outcome measure. As with T1, the timer start and stop were based on the (un)crossing of the wrists or when reliance on the safety tether was detected. The task was also capped at 30 seconds, as is the consensus recommendation[7].

Lastly, reactive balance was tested by applying two forward and two backward treadmill **perturbations (T5)** in random order, while participants stood with their feet together and were instructed to recover from the perturbation without taking a step. The treadmill applied the perturbation with: a 0.3 s acceleration phase – 0.5 s constant velocity phase – 0.3 s deceleration phase. The belt acceleration was set to 0.2 m/s^2 and the maximum velocity was 1.5 m/s (in the backward direction) and −1.0 m/s (in the forward direction). The configuration of the perturbation was based on Staring et al.[36] with tuning of the velocities based on protocol pilots with our target population[36]. Note that the direction of treadmill acceleration corresponds to imposed body sway in the opposite direction. A recovery was successful if the participant was able to keep standing without grabbing the handrail, taking a step, or leaning into the safety harness. The number of successful recoveries was used.

All tasks, except for walking, were executed twice per subset, so in total four times per condition, the average of which was used for further analysis. Before each set of tasks, the participants were guided through a short familiarization routine, to get used to the newly selected mode. This routine included independently moving the trunk in all directions, walking, and turning. An example of the narrow-based stance, tandem stance, and walking task can be seen in supplementary Movie S1. Written informed consent was obtained from all participants for the publication of any identifiable data, images, or videos included in this study.

After each task, the participants were asked to rate their experience of the balance assistive capabilities of the device via a 5-point Likert scale, where zero indicated no effect at all and five indicated a strong assistive effect. The score was subsequently averaged across conditions for each participant.

Measurement equipment

The experiments were performed on an instrumented treadmill (GRAIL, Motek B.V., NL) surrounded by 10 motion capture cameras (Vicon Motion Systems Ltd, UK). During all experiments, participants were secured by a safety harness to prevent falling. The Plug-In-Gait model was used for marker labeling and event detection was done using Nexus (Vicon Motion Systems Ltd, UK). Further processing and calculation of kinematic measures was done using MATLAB™ (MathWorks, Inc., USA).

Statistical analysis

For the analysis of the continuous outcome measures, a non-parametric Friedman test of differences among repeated measures was used to investigate condition effects, rendering the reported Chi-squared values. Significance level was set at $p < 0.05$. Subsequently, a Wilcoxon signed-rank test was conducted, and effect size was calculated, for each metric, comparing the individual differences between conditions. A Bonferroni correction was used to counteract the multiple comparisons problem. The Wilcoxon effect size (r) was also calculated, for which the interpretation values are: 0.10 to < 0.3 (small), 0.30 to < 0.5 (moderate) and ≥ 0.50 (large).

For the discrete perturbation outcome, a contingency table was made with the three conditions and the (non-)successful recovery counts. A Chi-square test was used to analyze differences between conditions.

As this is an explorative study sample size calculation was not performed. Statistical analysis was performed in R (R Foundation for Statistical Computing).

Data availability

All data and materials used in the analysis are available on DOI: 10.4121/c67bab10-e16a-4b53-a2f4-83fff48f9423. The raw Vicon and video data are available upon request to the corresponding author.

Code availability

All code used in the analysis are available on DOI: 10.4121/c67bab10-e16a-4b53-a2f4-83fff48f9423.

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Author contributions

J.N., L.V., V.W., K.P., B.S., and H.V. conceived the concept and methodological design. H.V., B.S., and K.P. designed and implemented the control architecture, B.W. and L.V. performed patient-screening and inclusion, and B.S. and L.V. conducted the experiments. B.S. conducted the data analysis. G.R., H.V., and J.N. directed, supervised and fully revised the research. All authors reviewed the manuscript.

Competing interests

The authors declare no competing interests.

Additional information

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