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The magnitude of external fall-inducing forces in subjects using the DreamMotion exoskeleton prototype in static body positions - a pilot study

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Abstract

Purpose

Exoskeleton robots are becoming increasingly popular due to improved robotic technologies and the positive perception of users. Lower limb exoskeletons are the most widely used as assistive devices for people with disabilities. The aim of the study was to determine the magnitude of forces necessary to induce the fall of a person using the Polish prototype of the exoskeleton robot.

Methods

Sixteen volunteers used DreamMotion prototype designed to perform medical tasks was tested. Measurements of the fall-inducing forces were performed in compliance with safety standards. Assessed were fall-inducing forces acting in various directions in 3 static, vertical body positions. In each test position, 10 trials were completed resulting in the effective measurement.

Results

In the 2-leg standing with posterior vector direction, the lowest value of fall-inducing force was recorded (mean 1.50kG). Also, in 1-leg standing position, the lowest value of the fall-inducing force was recorded with posterior vector direction (1.66kG). In the step position, the highest fall-inducing forces were recorded with the posterior (8.58 kG) and anterior (6.37kG) vector directions, the lowest – with the lateral vector direction towards the stepping limb (3.26kG).

Conclusions

The forces required to induce a fall in a person wearing the exoskeleton robot are relatively low, with relative forces ranging from 1.45% to 8.30% of the subject-ER setup weight. In both the 2-leg and 1-leg standing positions, the lowest fall-inducing forces were recorded when the force vector was directed posteriorly. The exoskeleton robot's design will likely need to be modified to enhance safety in this particular direction.

Keywords: safety, fall, force, exoskeleton robot

Introduction

The exoskeleton robot (ER) is defined as a programmable device that emulates human-like actions to accomplish human tasks (1). Nowadays, ERs are becoming increasingly popular due to improved robotic technologies and the positive perception of people towards interacting with robots. The application of the ER to the human body can be divided into three locations: (1) throughout the whole human body, (2) at the upper part of the body, i.e. the torso and arms, and (3) at the lower part of the body, i.e., from the waist down. Out of the types of ERs mentioned, lower limb exoskeletons are the most widely used as assistive devices for people with disabilities (2–5), and for purpose of power augmentation for military or industrial workers (6,7).

In all applications of ERs, the dynamic and static balance, prevention of falling, ensuring controller stability and smooth human-exoskeleton interaction are of critical importance for the safety of users (8–11). Two recent systematic reviews provide a compendium of knowledge with regard to the aspects mentioned above. Hamza et al. (8) review the advances in the falling recognition, balance recovery and stability assurance strategies in the design and application of ERs. It has been found that Zero Moment Point, Centre of Mass and Extrapolated Centre of Mass ideas are mostly used for balancing and prevention of falling. Sam et al. (9) indicated five most critical functions of ERs, including: (a) detection of fall, (b) estimation of user state, (c) estimation of user motion, (d) estimation of user intentional direction, and (e) detection of user balance loss. It is frequently the case that exact information on how to fulfil all these requirements comes from a special type of experiments including the controlled participants' falls and measurements of fall-inducing forces. They are crucial for advancing technologies aimed at preventing falls, especially in vulnerable populations such as people with neuro-motor deficits or in the elderly.

Fall-inducing experiments help researchers understand the physiological and mechanical responses involved in maintaining balance. By simulating conditions that lead to falls, it is possible to analyse how the body reacts and identify critical points where interventions, like ERs, can be most effective. For instance, Beck et al. emphasize that ERs must react faster than natural physiological responses to maintain standing equilibrium, highlighting the need for precise timing in robotic assistance (12). These experiments inform the design and functionality of ERs. By understanding how falls occur and which forces (their direction and magnitude) are most likely to initiate a fall, devices can be developed that anticipate and counteract these forces. Luo et al. take even a step further form extensive physical trials and present the potential of using simulations to develop exoskeleton assistance, thus accelerating innovation in the field (13). As falls are a leading cause of injury, their prevention constitutes also public health priority. Research indicates that effective fall prevention strategies can significantly reduce morbidity and mortality rates associated with falls. Gait planning in lower limb exoskeletons, which can be tailored to provide support during risky movements is a good example underlining this role of ERs in the society (14,15). Moreover, fall-inducing experiments also contribute to understanding user interactions with assistive devices. By studying how individuals respond to simulated falls while using exoskeletons, developers can enhance user experience and confidence in these technologies. This is vital for encouraging adoption of the ER as well as for overall mental condition among all individuals with disabilities who may fear falling (16,17).

When developing the prototype of the Polish ER DreamMotion, we emphasized the critical importance of user safety. The significant concerns surrounding stability and fall prevention led us to investigate the magnitude of forces necessary to induce fall of a person in the ER. In our experiment with induced falls assessed were forces acting in various directions relative to the subject's body (anterior, posterior, lateral) in 3 static, vertical body positions (2-

leg standing, 1-leg standing, step forward). The primary objective of this re-search was to gather insights that could inform necessary modifications to enhance the safety features of the existing ER prototype. By understanding how different forces interact with the human body in these scenarios, we aim to implement effective strategies that will significantly reduce fall risks in future iterations of the device.

Materials and Methods

The study received approval from the Ethics Committee of the Regional Medical Council in Krakow (72/KBL/OIL/2024) and was conducted in line with the Declaration of Helsinki. Participants were free to withdraw from the study at any time, though none chose to do so. All gave written informed consent to participate in the study.

Participants

In the presented study the purposive sample of volunteers was used. Each participant underwent a preliminary health screening conducted by a certified physiotherapist to ensure they met the inclusion criteria. The inclusion criteria were: no current pain within the musculoskeletal system, no history of orthopaedic or neurological problems resulting in long-term motor disability, no medical problems influencing body equilibrium, body mass index in the range of 23-25 kg/m^2, general good health condition, no minor medical maladies on the day of the examination (e.g. cold, headache, post-exercise muscle pain, etc.), fear of falling in the ER. Twenty-one people declared their willingness to take part in the research. Five volunteers were excluded due to too high body mass index (3 cases) and limited mobility after recent orthopaedic procedures (2 cases). Ultimately, 16 volunteers took part in the measurements. The basic characteristics of the study group are presented in Table 1.

Table 1. Basic demographic characteristics of the study group.

	mean ± std. dev. (min-max) or N (%)

gender	F = 6 (37.5); M = 10 (62.5)
age (years)	31.69 ± 5.39 (24.00-41.00)*
body mass (kg)	76.19 ± 7.55 (64.00-88.00)*
body and exoskeleton mass (kg)	103.19 ± 7.55 (91.00-115.00)*
body height (cm)	178.38 ± 7.84 (166.00-192.00)*
body mass index (kg/m^2)	23.88 ± 0.47 (23.18-24.81)*

*normal distribution according to Shapiro-Wilk test (P>0.05)

The exoskeleton and its comparison with similar devices

In the study examined were magnitudes of fall-inducing forces in a person using the Polish ER DreamMotion prototype designed to perform medical tasks. There are currently four similar exoskeletons on the market: REX (Rex Bionics, Melbourne, Australia)(18), ReWalk (Argo Medical Technologies Ltd, Marlborough, MA, USA)(19), Ekso GT (Ekso Bionics, San Rafael, CA, USA)(20), and Indego (Parker Hannifin, Macedonia, OH, USA)(21).

DreamMotion features four actively powered degrees of freedom at the hips and knees, similar to systems such as ReWalk, Ekso GT, and Indego. The ankle joint operates passively, without active control for balance compensation through the foot.

The control system in our prototype incorporates an absolute encoder, which provides precise, real-time data on joint positions and velocities, thereby enhancing both stability and motion control. The encoder works in conjunction with motor-reducers, enabling advanced control strategies such as dynamic balance compensation, smooth motion execution, and rapid response to destabilising forces. Indego employs a lean-based initiation mechanism, supported by a mobile application that provides user feedback. ReWalk utilises tilt sensors along with a wrist-worn remote control, relying on pre-programmed step triggers to initiate gait. REX is manually operated via a joystick. Ekso GT is predominantly therapist-controlled through an external interface.

In terms of user height compatibility, DreamMotion supports users between 150–190 cm, REX between 142–193 cm, Ekso GT between 157–188 cm, Indego between 157–190 cm, and ReWalk between 160–190 cm. Regarding ER's weight, Indego is the lightest at 12 kg, while REX is the heaviest at 38 kg. DreamMotion weighs 27 kg. The maximum supported user weight for DreamMotion is 90 kg, while ReWalk, Ekso GT, and REX accommodate up to 100 kg, and Indego supports up to 113 kg.

DreamMotion provides up to 8 hours of operation per charge, similar to ReWalk. Indego and Ekso GT offer approximately 4 hours, while REX provides around 2 hours.

In terms of walking speed, DreamMotion reaches up to 3 km/h. ReWalk achieves approximately 2.3–2.5 km/h, Ekso GT around 1.6 km/h, Indego ranges from 1.1–2.4 km/h, and REX is designed for slower movement, with a speed of approximately 1 km/h.

Research team

The research team included an engineer participating in the construction process of the ER prototype, responsible for adapting the device to the size of the subject's body and adjusting it for demands of each test position (see below), as well as controlling and eliminating any technical disruptions occurring during the measurements. The physiotherapist supported the process of setting up the ER, was responsible for positioning the subject in the measurement station, installing safety systems, and providing additional protection during fall trials. The laboratory technician operated devices applying forces, recorded and extracted data, and prepared the database for statistical calculations. He was the only person who did not know the purpose of the study. The research supervisor, not directly involved in the measurements, was responsible for recruiting the subjects and conducting the statistical analysis of the results. All members of the research team underwent appropriate training during the pilot studies. After their completion, they presented appropriate competences and operational efficiency.

Measurement of force magnitude

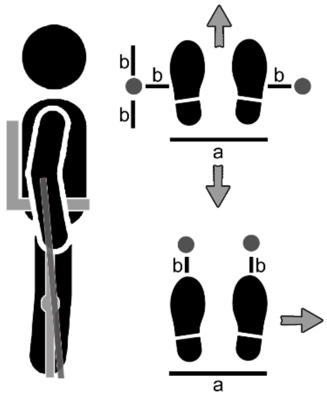
Measurements of fall-inducing forces in a person using the ER were carried out in a motion analysis lab, following all safety standards. Each subject first put on a full-body mountaineering harness with a hip and shoulder belt, then stepped into the ER. The ER was then adjusted to fit the subject's body. The thigh and shin sections were set to match the person's leg lengths, and the ER hinges were aligned with the medial-lateral axes of the hip and knee joints. This ensured both comfort and proper function. The subject was positioned at a measurement station with a ceiling-mounted suspension system for safety. Six elastic ropes (8 mm diameter) were attached to the harness. These could fully support the subject's weight along with the equipment. The ropes were anchored to the ceiling at two points, 1 metre apart. A backup safety system, i.e. a single static rope (8 mm diameter) was also connected to the harness and anchored to one ceiling point. The elastic ropes were set to stop a fall after 20-30-cm "flight". The static rope would catch the subject slightly later, but was never actually needed. A third safety measure was a physiotherapist from the research team. They stood close by during each trial, ready to intervene manually if anything unexpected happened, such as a sudden body rotation during a fall.

In all test positions, the fall-inducing force was always applied at shoulder height, either perpendicular to the frontal plane of the body (anterior or posterior force vector direction) or perpendicular to the sagittal plane (lateral force vector direction (right or left)). An additional static rope (8 mm in diameter) was used for this purpose. For each force vector direction, the rope was attached to different harness elements located in the area of: sternal notch (anterior direction), second thoracic spinous process (posterior direction), right or left acromion (right or left direction, respectively). The rope was connected to an external sensor (DEE 750 kG, Keli Sensing Technology, Ningbo, China) of the AxisFB50 electronic dynamometer (Axis, Gdańsk, Poland) with a maximum capacity of 500 kG and a measurement accuracy of 0.01 kG. The sensor was connected with a steel rope to the Vevor PA300 mini electric winch (Vevor, Shanghai, China) with a maximum load capacity of 300 kG and power of 550 W. A constant rope movement speed of 0.08 m/s was used. The order of the test positions and directions of the applied forces were randomized for each participant to minimize the effects of fatigue and learning. All measurement devices were calibrated before each testing session according to the manufacturer's instructions.

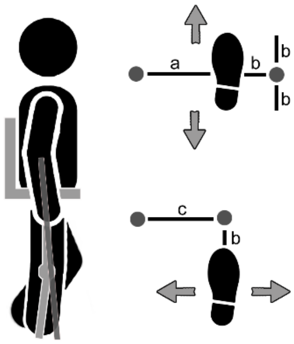
During the pilot studies, all components of the measurement station were assembled and tested. Suitable subject-ER setup positioning positions that allowed for effective were identified. Three positions were selected with the following directions of fall-inducing force vectors:

- symmetrical 2-leg standing position (ER hinges set in their zero position corresponding to the anatomical position of the body) – 1) anterior direction of the force vector, 2) posterior direction, 3) lateral direction to the right;
- 1-leg standing position on the right limb (joints of the left ER’s "limb" set in the position: hip joint – flexion 20°, knee joint – flexion 20°; joints of the right ER’s "limb" set in the anatomical position) – 4) anterior direction of the force vector, 5) posterior direction, 6) lateral direction towards the supporting limb (right), 7) lateral direction towards the raised limb (left);
- step position with the right limb (joints of the left ER’s "limb" set in the position: hip joint – 5° flexion, knee joint – 5° flexion; joints of the right ER’s "limb" set in the position: hip joint – 20° flexion, knee joint – flexion 10°) – 8) anterior direction of the force vector, 9) posterior direction, 10) lateral direction towards the stepping limb (right), 11) lateral direction towards the supporting limb (left).

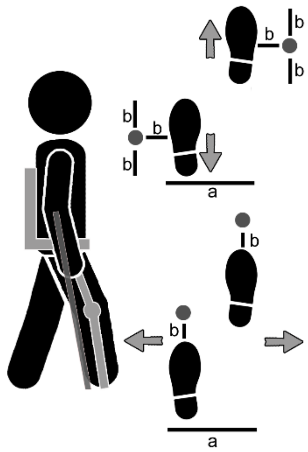
The indicated ER settings were dictated by the ability to maintain balance and the comfort of the subject. In each of these positions, elbow crutches were also used. They were adjusted so that the flexion angle in the subject's elbow did not exceed 10°. The crutches were placed on the ground in such a way so they could provide support, but did not widen the support polygon in the direction of the fall-inducing force (broadening the polygon in this direction would allow a strong person to completely support their body weight on the crutch and stop the fall; such a situation would make it impossible to achieve the study goals). Wearing the ER with its joints locked in the positions described above, and using elbow crutches, undoubtedly limited the participants' ability to perform natural balance adjustments. However, this setup was deliberately designed to simulate the conditions faced by the intended users of the ER –individuals with disabilities that prevent movement of the lower limbs. As the ER's joints were locked and no movement was possible, we chose not to control for the participants' dominant lower limb. The one-leg standing and step positions were therefore tested only with weight borne on the right leg and the right leg used for swinging, respectively. Diagrams of the subsequent test positions and the arrangement of the feet and crutches on the floor are presented in Figure 1.



left – body arrangement in test positions 1, 2, 3
right up – feet and crutches (grey dots) arrangement in test positions 1, 2
right down – feet and crutches (grey dots) arrangement in test position 3
a – 30 cm
b – ½ foot length
grey arrows – force vectors applied in the given feet/crutches arrangement
elbows – max 10° flexion
hips, knees – 0°



left – body arrangement in test positions 4, 5, 6, 7
right up – feet and crutches (grey dots) arrangement in test positions 4, 5
right down – feet and crutches (grey dots) arrangement in test position 6, 7
a – 20 cm + ½ foot length
b – ½ foot length
c – 20 cm
grey arrows – force vectors applied in the given feet/crutches arrangement
elbows – max 10° flexion
hips: right – 0°, left – 20° flexion
knees: right – 0°; left – 20° flexion



left – body arrangement in test positions 8, 9, 10, 11
right up – feet and crutches (grey dots) arrangement in test positions 8, 9
right down – feet and crutches (grey dots) arrangement in test position 10, 11
a – 30 cm
b – ½ foot length
grey arrows – force vectors applied in the given feet/crutches arrangement
elbows – max 10° flexion
hips: right – 20° flexion, left – 5° extension
knees: right – 10° flexion; left – 10° flexion

Figure 1. Schematic presentation of the consecutive test positions and fall-inducing force vector direction together with the arrangement of the feet and crutches on the floor. Top: 2-leg standing position: 1 – posterior; 2 – anterior; and 3 – lateral (right) vector direction; middle: 1-leg standing position (right leg supporting): 4 – posterior; 5 – anterior; 6 – lateral right; and 7 – lateral left; bottom: step position (right leg swinging): 8 – posterior; 9 – anterior; 10 – lateral right; 11 – lateral left vector direction.

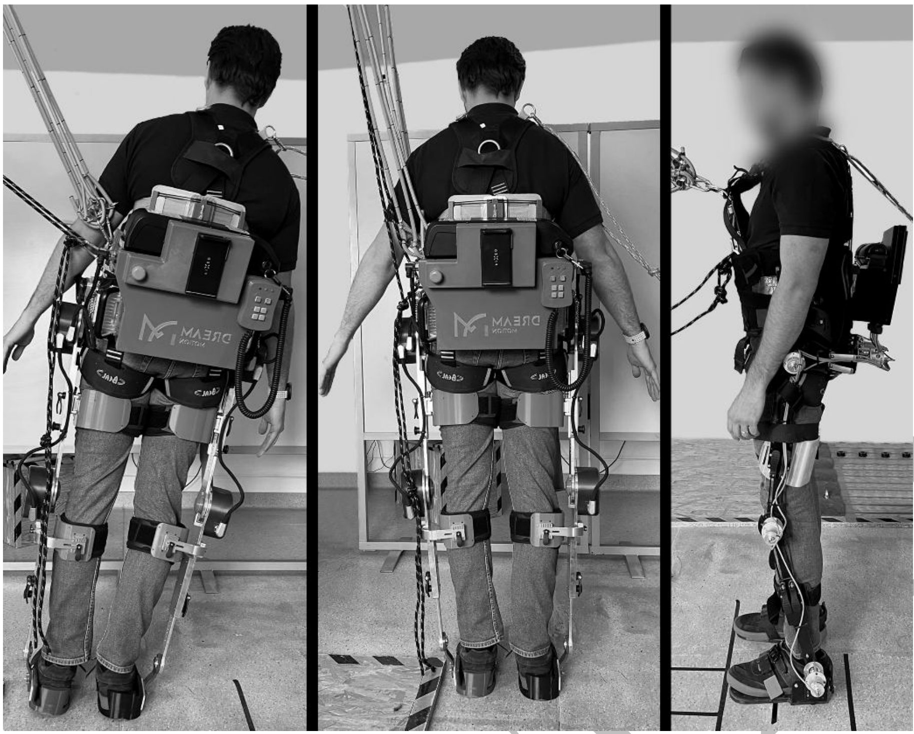


Figure 2. Examples of the subject-exoskeleton setup positions in the measurement station. To preserve clarity crutches are not presented. Left: final position after the "flight" started in the 2-leg standing position with lateral (right) direction of the fall-inducing force vector; safety system ropes are located on the left side of the participant's body. Middle: initial 2-leg standing position with lateral (right) force vector direction. Right: initial 2-leg standing position with posterior force vector direction; safety system ropes are located in the front of the participant's body.

In each test position, the subject's task was to maintain body balance as long as possible with their eyes closed. The subject could not use crutches for this purpose, i.e. change their position on the ground. Before applying the fall-inducing force, the application rope was kept in a state of pre-load of up to 0.1 kG, and then the electric winch was started to introduce the true fall-inducing force. The time interval between the preload and the true force was adjusted randomly in the range from 3 to 10 s to eliminate the effect of the subject's preparation for the application of the true force. A fall was considered complete when the subject lost their balance and hung freely in the suspension system. If the crutch was moved on the ground, the test was repeated. In each test position, 10 trials were completed resulting in the effective measurement of the peak value of the fall-inducing force. Examples of the subject-ER setup positions in the measurement station are presented in Figure 2.

Procedure

After completing the recruitment and verification of the inclusion criteria, qualified volunteers were offered a convenient date to visit the motion analysis laboratory and complete the procedure. All participants were required to wear non-restrictive clothing and soft shoes. A physiotherapist and an engineer from the research team explained the details of the further part of the procedure to the subjects. Subsequently, the subjects were placed in the ER, took their position in the measurement station and all the safety systems were installed. In order to get familiar with the sensations generated during controlled falls, the subject performed several test falls, initially at a slow pace with manual support from the physiotherapist and engineer. Next, there were several test falls without manual assistance from team members. When the subject declared sufficient readiness to start the measurements, the actual procedure was started. The application rope was connected to the harness and the fall-inducing force was generated, controlled by the technical assistant. In each test position, he commanded the subject: "Your task is to maintain balance as long as you can. Ready?" After receiving the answer: "Yes!", he first applied the pre-load and then, at an interval of 3-10 s, the true fall-inducing force. In case of problems (e.g. displacement of the crutches on the floor), the test was repeated. In each test position, 10 measurements were taken. The peak forces were recorded in the dynamometer memory and the technical assistant remained unaware of their values until the measurements were completed for a given person. Then, he transmitted the data to the computer disk where the database was created.

The order in which the different positions and force vectors directions were tested was semi-random. Due to time economy, only the positions were randomized (2-leg standing, 1-leg standing, step), while the order of the vector directions remained constant: anterior, posterior, lateral (in 1-leg standing and step positions: 1) towards the supporting/stepping limb; 2) towards the raised/supporting limb, respectively). The entire procedure took about 1.5 hours.

Data processing

The applied fall-inducing force increased gradually, reached its peak at the moment the fall was initiated, then dropped sharply. These peak values can be interpreted as the minimum force required to induce a fall. For each participant, peak values were recorded across 10 trials in every test position and for each force vector direction. The mean of these 10 values was then calculated and used as the outcome measure for each position and force vector direction (hereafter referred to as absolute force).

The calculated absolute forces were also normalised to the total weight of the subject-ER setup and expressed as a percentage (hereafter referred to as relative forces).

For each set of 10 trials, the individual range of fall-inducing forces was calculated by subtracting the minimum from the maximum recorded force. This range was expressed as the absolute value of the difference.

As a result, three dependent variables were subjected to statistical analysis: absolute force magnitude, relative force magnitude, and intra-individual force range – each analysed for every test position and force vector direction.

Statistical analysis

During the statistical analysis, mean values, standard deviations, and minimum and maximum values of all dependent variables were calculated across all 16 participants. Differences in the dependent variables between test positions and between directions of the force vector were assessed using the standard Student's t-test for dependent data, applied across all participants. The critical *P* level was set at 0.05. Statistica 13.0 PL software (StatSoft, Tulsa, USA) was used.

Results

The fall-inducing forces magnitude were generally small. In absolute values, they reached a maximum of about 10 kG (Figure 4), while after normalization to the subject-ER setup weight – a maximum of about 9% of this weight (Figure 5). However, these forces showed significant variation in different initial positions and with different directions force vectors. In the 2-leg standing and posterior vector direction, the lowest value of fall-inducing force was recorded (position 1; mean absolute force 1.50 kG, mean relative force 1.45%). In the anterior direction (position 2), these values were the highest (mean absolute force 4.18 kG, mean relative force 4.05%). Also in the 1-leg standing position, the lowest values of the fall-inducing forces were recorded with posterior vector direction (position 4; mean absolute force 1.66 kG, mean relative force 1.60%). In the step position, a different situation occurred. The highest fall-inducing forces were recorded with the posterior (position 8; mean absolute force 8.58 kG, mean relative force 8.30%) and anterior (position 9; mean absolute force 6.37 kG, mean relative force 6.17%) vector directions, the lowest – with the lateral vector direction (position 10 towards the supporting limb; mean absolute force 2.99 kG, mean relative force 2.89%; position 11 towards the stepping limb; mean absolute force 3.26 kG, mean relative force 3.16%). The detailed data on forces recorded in all test positions are presented in Table 2.

Table 2. Mean values ± standard deviations (minima-maxima) of the absolute forces, relative forces and intra-individual ranges of forces causing falls in consecutive test positions and [vector directions]. 1 – 2-leg standing [posterior]; 2 – 2-leg standing [anterior]; 3 – 2-leg standing [lateral (right)]; 4 – 1-leg standing (right leg supporting) [posterior]; 5 – 1-leg standing (right leg supporting) [anterior]; 6 – 1-leg standing (right leg supporting) [lateral right]; 7 – 1-leg standing (right leg supporting) [lateral left]; 8 – step (right leg swinging) [posterior]; 9 – step (right leg swinging) [anterior]; 10 – step (right leg swinging) [lateral right]; 11 – step (right leg swinging) [lateral left].

position	absolute force (kG)	relative force (%)	range (kG)
1	1.50 ± 0.16 (1.29-1.81)	1.45 ± 0.06 (1.38-1.58)	0.93 ± 0.10 (0.80-1.13)
2	4.18 ± 0.44 (3.60-5.05)	4.05 ± 0.17 (3.86-4.39)	1.03 ± 0.11 (0.88-1.24)
3	2.74 ± 0.29 (2.35-3.31)	2.65 ± 0.11 (2.53-2.88)	1.96 ± 0.21 (1.68-2.36)
4	1.66 ± 0.24 (1.34-2.12)	1.60 ± 0.12 (1.45-1.85)	1.58 ± 0.17 (1.36-1.91)
5	2.94 ± 0.31 (2.52-3.55)	2.84 ± 0.12 (2.71-3.08)	1.77 ± 0.19 (1.52-2.14)
6	2.96 ± 0.31 (2.55-3.58)	2.87 ± 0.12 (2.73-3.11)	1.03 ± 0.11 (0.88-1.24)
7	2.66 ± 0.28 (2.28-3.21)	2.57 ± 0.11 (2.45-2.79)	2.05 ± 0.22 (1.76-2.48)
8	8.58 ± 0.91 (7.38-10.36)	8.30 ± 0.34 (7.92-9.01)	1.40 ± 0.15 (1.20-1.69)
9	6.37 ± 0.67 (5.48-7.70)	6.17 ± 0.25 (5.88-6.69)	2.98 ± 0.32 (2.56-3.60)
10	2.99 ± 0.32 (2.57-3.61)	2.89 ± 0.12 (2.76-3.14)	1.68 ± 0.18 (1.44-2.03)

11	3.26 ± 0.35 (2.80-3.94)	3.16 ± 0.13 (3.01-3.43)	1.86 ± 0.20 (1.60-2.25)

In most cases, significant differences were recorded of the absolute fall-inducing forces between the subsequent test positions. There were only a few exceptions to this tendency. The matrix of *P* values from the Student's t-test is presented in Table 3.

Table 3. Matrix o *P*-values (t-Student test for the dependent data) in the analysis of differences between the consecutive test positions and [vector directions] for the parameter: absolute force magnitude. 1 – 2-leg standing [posterior]; 2 – 2-leg standing [anterior]; 3 – 2-leg standing [lateral (right)]; 4 – 1-leg standing (right leg supporting) [posterior]; 5 – 1-leg standing (right leg supporting) [anterior]; 6 – 1-leg standing (right leg supporting) [lateral right]; 7 – 1-leg standing (right leg supporting) [lateral left]; 8 – step (right leg swinging) [posterior]; 9 – step (right leg swinging) [anterior]; 10 – step (right leg swinging) [lateral right]; 11 – step (right leg swinging) [lateral left].

position	1 1.50 kG	2 4.18 kG	3 2.74 kG	4 1.66 kG	5 2.94 kG	6 2.96 kG	7 2.66 kG	8 8.58 kG	9 6.37 kG	10 2.99 kG
2	<0.001									
3	<0.001	<0.001								
4	0.635	<0.001	<0.001							
5	<0.001	<0.001	0.274	<0.001						
6	<0.001	<0.001	0.117	<0.001	0.997					
7	<0.001	<0.001	0.991	<0.001	<0.05	<0.05				
8	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001			
9	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001		
10	<0.001	<0.001	<0.05	<0.001	0.714	0.997	<0.01	<0.001	<0.001	
11	<0.001	<0.001	<0.001	<0.001	<0.001	<0.05	<0.001	<0.001	<0.001	<0.05

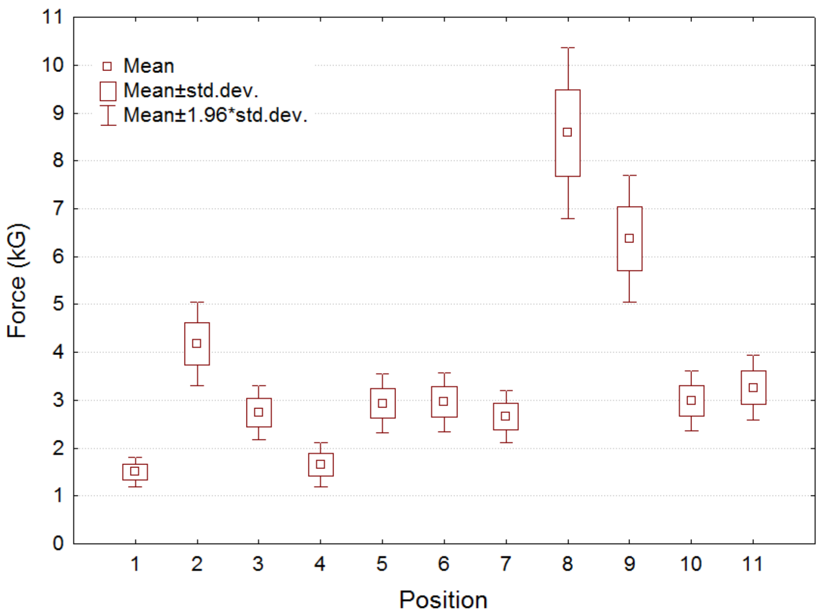


Figure 3. Mean absolute values and standard deviations of the forces causing falls in consecutive test positions and [vector directions]. 1 – 2-leg standing [posterior]; 2 – 2-leg standing [anterior]; 3 – 2-leg standing [lateral (right)]; 4 – 1-leg standing (right leg supporting) [posterior]; 5 – 1-leg standing (right leg supporting) [anterior]; 6 – 1-leg standing (right leg supporting) [lateral right]; 7 – 1-leg standing (right leg supporting) [lateral left]; 8 – step (right leg swinging) [posterior]; 9 – step (right leg swinging) [anterior]; 10 – step (right leg swinging) [lateral right]; 11 – step (right leg swinging) [lateral left].

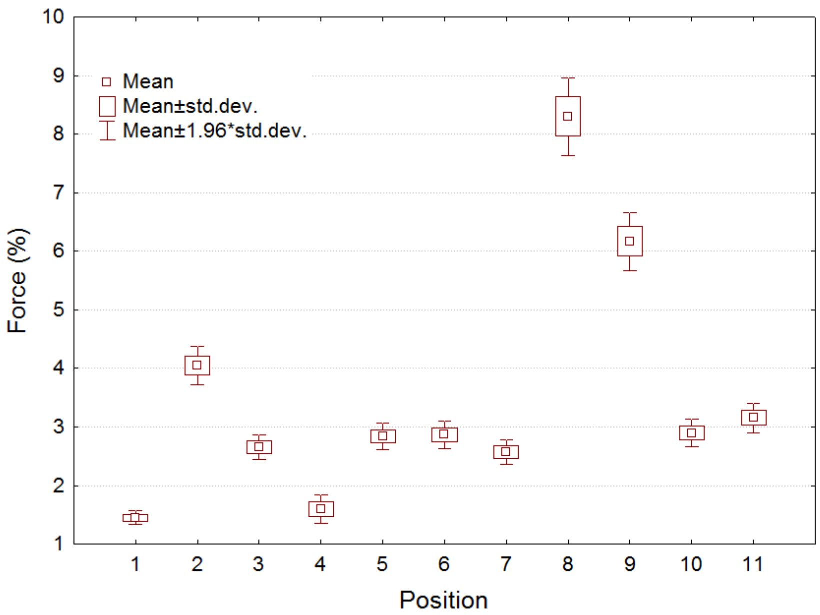


Figure 4. Mean relative values (normalized for the body plus exoskeleton weight) and standard deviations of the forces causing falls in consecutive test positions and [vector directions]. 1 – 2-leg standing [posterior]; 2 – 2-leg standing [anterior]; 3 – 2-leg standing [lateral (right)]; 4 – 1-leg standing (right leg supporting) [posterior]; 5 – 1-leg standing (right leg supporting) [anterior]; 6 – 1-leg standing (right leg supporting) [lateral right]; 7 – 1-leg standing (right leg supporting) [lateral left]; 8 – step (right leg swinging) [posterior]; 9 – step (right leg swinging) [anterior]; 10 – step (right leg swinging) [lateral right]; 11 – step (right leg swinging) [lateral left].

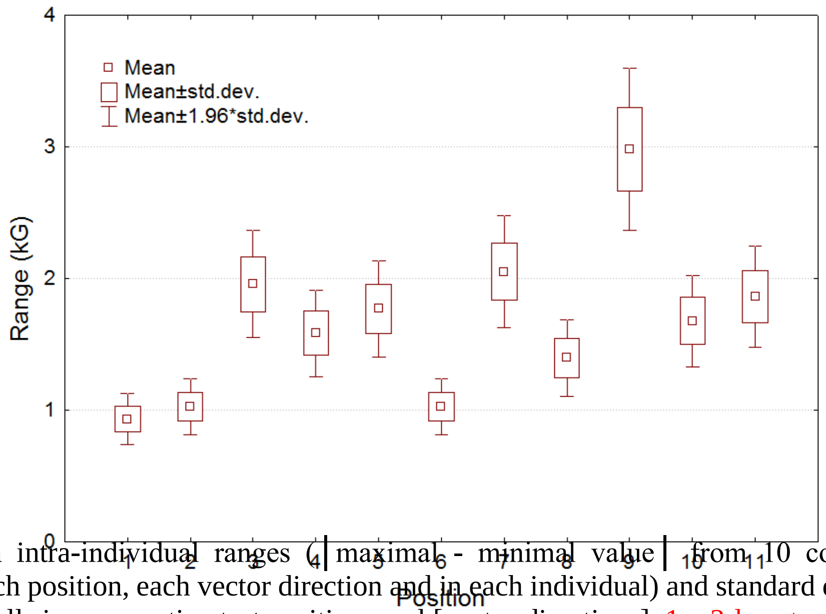


Figure 5. Mean intra-individual ranges (|maximal - minimal value| from 10 consecutive trials performed in each position, each vector direction and in each individual) and standard deviations of the forces causing falls in consecutive test positions and [vector directions]. 1 – 2-leg standing [posterior]; 2 – 2-leg standing [anterior]; 3 – 2-leg standing [lateral (right)]; 4 – 1-leg standing (right leg supporting) [posterior]; 5 – 1-leg standing (right leg supporting) [anterior]; 6 – 1-leg standing (right leg supporting)

[lateral right]; 7 – 1-leg standing (right leg supporting) [lateral left]; 8 – step (right leg swinging) [posterior]; 9 – step (right leg swinging) [anterior]; 10 – step (right leg swinging) [lateral right]; 11 – step (right leg swinging) [lateral left].

The ranges of the fall-inducing forces showed certain intra-individual variability during the 10 trials executed in each test position (Figure 6). The smallest ranges, with the mean value of about 1 kG, were recorded in the 2-leg stance with the anterior and posterior vector directions, as well as in the 1-leg stance with the lateral vector direction towards the supporting limb. The largest range, with a mean size of about 3 kG, occurred in the step position with the anterior vector direction. Detailed data are presented in Table 2.

Discussion

In recent years, the demand and usage of ERs are gradually increasing in diverse areas. The main applications of ER are assistance, rehabilitation and power augmentation (8). This trend imposes the obligation to ensure high user safety on designers and manufacturers. In this aspect, many research have been undertaken, in which the development of stability ensuring systems and protection against falls were of special interest (8, 22-24, 28-29).

Khalili et al. (22) developed optimization techniques for safe fall strategies in lower limb ERs, reducing hip impact velocity by over 50%. The study was motivated by the need to address the safety concerns that currently limit the independent use of lower limb exoskeletons, as no control strategy has yet been implemented that prevents falls in the case of a loss of balance. Hamza et al. (8) reviewed balance and stability issues, highlighting the use of Zero Moment Point, Center of Mass, and Extrapolated Center of Mass concepts for fall prevention. Crea et al. (23) emphasized the importance of field validation studies for large-scale adoption of occupational ERs, proposing a roadmap to facilitate informed decision-making among stakeholders. The analysis of the state-of-the-art shows methodological differences between laboratory and field studies. While the former are more extensively reported in scientific papers, they exhibit limited generalizability of the findings to real-world scenarios. On the contrary, field studies are limited in sample sizes and frequently focused only on subjective metrics. Mahdian et al. (24) advocated for incorporating muscle biomechanics principles in exoskeleton design and control, suggesting the development of predictive controllers that optimize both biological and electromechanical performance. These studies collectively underscore the need for integrating biomechanical considerations into exoskeleton design to enhance safety, stability, and user adaptation. Real-time mapping of the neuromechanical origin and generation of muscle force resulting in joint torques should be combined with musculoskeletal models to address time-varying parameters such as adaptation to ERs and fatigue. Development of smarter predictive controllers that steer rather than assist biological components could result in a synchronized human-machine system that optimizes the biological and electromechanical performance of the combined system.

Stability is one of the crucial concerns for the current ERs. Sophisticated control strategies or supplementary external supports are often required to achieve stability (25). In so-called static approaches, the vertical projection of the centre of mass is strictly limited to the inside of the support polygon (26). In dynamic approaches, the projected centre of mass may leave the support polygon (27). Our research was conducted in static conditions, however our intention was to displace the centre of mass projection outside of the support polygon and induce a controlled fall. The participants were healthy volunteers, which is a common practice in this type of measurement (to avoid exposure of subjects with disabilities to fear and discomfort related to the procedures)(28–31). The results clearly indicate that very small forces are sufficient to induce a fall of a person using ER, reaching at best a mean value of 8.30% of the subject-ER setup weight (posterior vector direction in the step position) when acting at the height of the subject's shoulders. In the 2-leg standing position, even the forces of a mean magnitude of 1.45% of the subject-ER setup weight (posterior vector direction) are sufficient to induce a fall. Such low values were obtained despite the fact that the subjects always used elbow crutches. Statistical analysis suggests that the differences in fall-inducing force magnitude across the examined positions and force vector directions can largely be generalised to the wider population (Table 3).

Some observations related to the direction of a fall in dynamic conditions (in opposition to ours) were described by Tan et al. (32). These authors measured and compared the distances between ER users and their four wheeled walkers (4WWs) during level and slope walking. The distances increased in uphill slope conditions and decreased in downhill slope conditions. Authors concluded that changes in the distance between the ER user and the walker may lead to an increase in the risk of falling forward on an uphill slope and backward on a downhill slope, as compared to a level surface. The risk of falling might be higher on the downhill slope condition because the most frequent unexpected postural disturbances occurred in the posterior direction caused by the short distance between the ER user and the walker.

In our study, the posterior direction of the fall-inducing force vector forces turned out to be the most critical, too, both in the 2-leg and 1-leg standing positions. This was probably due to the fact that even in a healthy person, the posterior margin of stability in an upright standing position is always the narrowest one. This situation was additionally complicated in the case of using ER by equipping the device with a kind of 'backpack' containing batteries and electronic components. Its location caused the overall centre of gravity of the subject-ER setup to rise (leading to a reduction of the angles of stability) and to move posteriorly (further reduction of the posterior margin of stability), which created the most favourable (as compared to the other vector directions) preconditions for a fall. Shin et al. (33) also point out that external load affects the displacement of the centre of mass, which may consequently cause the user to fall, especially on an inclined or uneven surfaces.

Interestingly, adopting the 1-leg standing position did not significantly worsen the situation with the posterior and lateral vector directions. The influence of this position was revealed only with the anterior direction of the vector. Assuming the step position caused the subject-ER setup to become relatively more stable in the sagittal plane. However, it was significantly easier to induce a fall in this position by applying a lateral force. The lateral vector direction was found to be critical not only in our study. Ramanujam et al. (28) assessed bi-planar dynamic stability margins for healthy adults during robot-assisted walking using EksoGT, ReWalk, and Indego compared to independent overground walking at slow, self-selected, and fast speeds. Despite the dissimilarities in the design and operation of these ER devices, the dynamic margins of stability for these individuals were found to be lower during self-selected speed, especially in the medial-lateral direction across all devices. Similar circumstances occur also during turns, when the user assumes the position of so called tandem stance. In this position both legs form a line along walking direction, the support polygon narrows and elongates, and lateral stability margins shrink (29). The lateral force vectors, reduced margins of stability in the medial-lateral direction and assuming the tandem position, all these factors seem to constitute a favourable conditions for lateral falls.

The frequently recorded significant differences in the magnitude of fall-inducing forces in different test positions seem to have scientific rather than practical importance. The forces themselves were small, and so were the differences between them. Substantively significant can be considered: 1) the differences in the magnitude of forces in the posterior direction in the 1-leg and 2-leg standing position from all other forces; 2) the differences in the magnitude of forces in the posterior and anterior directions in the step position from all other forces.

The obtained results prompt several reflections. Firstly, the recorded image indicates that the ER users will be forced to support themselves in an upright standing position with crutches, which, of course, will limit their ability to use their upper limbs for other activities.

Secondly, the most susceptible to induce falls seem to be the posterior force in the 1-leg and 2-leg standing positions, and the lateral forces in the step position. This introduces the need to introduce additional ER safeguards to protect the user from falling backwards and sideways. It will probably be necessary to widen the ER's support polygon in these directions, but the method of achieving this effect has yet to emerge from the conceptual sphere. It is worth emphasising that most ERs approved for use require increased stability through the introduction of additional aids such as bilateral canes, forearm crutches. However, the use of such aids does not guarantee full safety. Slipping or sliding of the crutch tip due to the material used or character of the walking surface (e.g., wet pavements, snow, ice, etc.) can lead to fall. Only a few exoskeletons have been constructed that are able to provide the compete fall protection – all have a walker structure (e.g. ATALANTE, ATLAS, MINDWALKER, Hyundai Medical Exoskeleton (H-MEX))(8, 33-34). All types have gait assistance in both sagittal and lateral planes and free upper extremities.

Limitations of the Study

This study has several notable limitations. It was conducted on a small sample of 16 healthy volunteers. While our findings can largely be generalised to populations with similar characteristics (Table 3), their applicability to individuals with disabilities is limited. Measurements were taken in static body positions, which do not fully reflect the dynamic conditions of everyday exoskeleton use. Participants used elbow crutches, which may have influenced the results, and the laboratory conditions differ from real-world environments. Additionally, the study did not include a long-term evaluation of exoskeleton use, which is crucial for fully understanding user adaptation to the device. Despite these limitations, the study provides valuable preliminary data for further research.

Conclusions

The forces required to induce a fall in the ER were generally small, with relative mean values ranging from 1.45% to 8.30% of the subject-ER setup weight. The study identified that the posterior direction of applied force, particularly in the 2-leg and 1-leg standing positions, was the most critical for stability, as it required the smallest forces to induce a fall. Additionally, lateral forces, especially in the step position, also posed a significant risk. These findings highlight the need for design modifications to improve stability in the posterior and lateral directions.

To reduce fall risks, future exoskeleton designs should focus on enhancing stability in these critical areas, potentially by widening the support polygon or incorporating additional safety features. Improved control systems that can respond to changes in posture and external forces in real-time could also play a key role in minimizing falls. These changes, along with long-term studies involving users with disabilities, will be essential to ensure safer and more reliable exoskeleton use in dynamic, real-world environments. Further tests on larger sample sizes, particularly in dynamic conditions, are planned. These will require significant modifications to the instrumentation to better simulate real-world usage and capture the broader range of forces and movements encountered in daily activities.

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