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Neuromuscular diagnostics and sensorimotor performance in training and therapy - beyond the pure biomechanical approach

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Neuromuscular diagnostics and sensorimotor performance in training and therapy - beyond the pure biomechanical approach

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Editorial: Neuromuscular diagnostics and sensorimotor performance in training and therapy - beyond the pure biomechanical approach

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Editorial on the Research Topic
Neuromuscular diagnostics and sensorimotor performance in training and therapy - beyond the pure biomechanical approach

Traditional movement science research has adopted methodology that describes differences in movement among groups or conditions using biomechanical variables to infer underlying features of neuromuscular control. Historically, this approach marked the onset of the complex analysis of movement biomechanics bring relevant insights into the mechanics of human movement (1).

Hypotheses were developed by extrapolating data from healthy active to injured populations (2). The last decade before the millennium provided first references to the neuromechanical aspects of movement, thus expanding the view towards the neuromotor control aspect of movement generation and adjustment (3). In the 2000s, substantial evidence is reported on spinal and supraspinal adaptations after balance or sensorimotor training which delivered important evidence-based knowledge that was rapidly implemented in clinical practice (4). Moreover, studies that combined both “mechanical” and “neuromuscular” views evolved (5, 6). We can postulate, that we still need more evidence-based knowledge on the interplay between the underlying neurophysiologic movement generation and the observed mechanical motor output. These integral neuro-biomechanical approaches still rely heavily on a biomedical perspective that is lately challenged by the call for biopsychosocial paradigms to cover all relevant aspects in human movement analysis to draw meaningful conclusions for diagnostics, prevention and therapy (7).

Research can rarely incorporate all dimensions at one time but our claim should be that we focus on experimental paradigms that purposely integrate both biomechanical and neuromechanical pieces of the puzzle to seek a more comprehensive understanding of typical and impaired movement. There are promising examples of such approaches that now combine classic biomechanical research with neurophysiological methods and patient reported outcomes or other psychometric measures (8, 9).

The aim of this Research Topic is therefore to provide a collection of studies that contribute to these integrative approaches by using diverse viewpoints and subsequently diverse methodology from study protocols, scoping or systematic reviews or experimental and interventional studies. They all contribute with different pieces of the puzzle "beyond the pure biomechanical approach".

Three investigations provided insight into motor control and muscle coordination in patient populations and those with experimentally imposed pian. Bartsch-Jimenez et al. described differences in "fine synergies" derived from electromyographic data of multiple lower leg muscles between persons with foot drop and controls that may reflect potentially relevant for motor adaptations to impaired ankle control. Chan and Sigward found that achieving loading symmetry in standing requires attention in those who are recovering from ACL reconstruction while it is more automatic in healthy controls. Bertrand-Charette et al. described the influence of acute ankle pain on motor output and performance of a standard balance test used to assess function in individuals with ankle injuries. While these studies targeted specific adaptations, Quarmby et al.'s systematic review of evidence regarding mechanical and neuromuscular control impairments in individuals with Achilles tendinopathy highlights limited consensus and areas for future work.

Other contributors provided insight into the effects of neurocognitive and neurophysiological based interventions. Rogan and Taeymans describe in their systematic review the evidence of positive effects of whole-body vibration on sensorimotor function in the elderly which highlights the therapeutic potential in this population. Faes et al. investigated the effects of a whole-body vibration intervention on several dimensions like movement control, well-being, and cognition in a randomized controlled trial. Hegi et al. summarized the existing body of evidence on sensor-based augmented visual feedback that should be used in coordination training to elicit sensorimotor adaptations. Mourits et al. describe a study protocol of a quasi-randomized controlled trial investigation of a game based intervention that combines neurocognitive effects of an external focus of attention and game like motivation along with patient specific real time spine motion to improve movement control of the spine. Finally, Mathieu-Kälin et al. described an assessment tool for develop to measure movement quality during hop tests. This tool adds important valuation of the control strategies used to complete a task beyond that of just performance.

The goal of the Research Topic was accomplished by presenting studies that incorporated a variety of manscirpt that represent "out of the box" neuro-biomechanical approaches to investigate underlying features of impaired movement. The broad range of paradigms and methodological approaches of the Research Topic certainly reflects the initial idea and the contributions highlight different aspects on the pathway to more mutifaceted approaches.

The guest editor team would love to see many views, downloads, and citations of the papers included in this Research Topic and we anticipate that in the future more contributions to Frontiers and Sports and Active living could be "virtually" added to this topic.

Author contributions

HB: Conceptualization, Supervision, Writing – original draft, Writing – review & editing. BP: Conceptualization, Supervision, Writing – review & editing. SS: Conceptualization, Supervision, Writing – review & editing.

Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Sensor-based intervention to enhance movement control of the spine in low back pain: Protocol for a quasi-randomized controlled trial

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Introduction: Chronic low back pain is a common condition that imposes an enormous burden on individuals and society. Physical exercise with education is the most effective treatment, but generally results in small, albeit significant improvements. However, which type of exercise is most effective remains unknown. Core stability training is often used to improve muscle strength and spinal stability in these patients. The majority of the core stability exercises mentioned in intervention studies involve no spinal movements (static motor control exercises). It is questionable if these exercises would improve controlled movements of the spine. Sensor-based exergames controlled with spinal movements could help improve movement control of the spine. The primary aim of this study is to compare the effects of such sensor-based exergames to static motor control exercises on spinal movement control.

Methods and analysis: In this quasi-randomized controlled trial, 60 patients with chronic low back pain who are already enrolled in a multidisciplinary rehabilitation programme will be recruited. Patients will be randomly allocated into one of two groups: the Sensor-Based Movement Control group ($n = 30$) or the Static Motor Control group ($n = 30$). Both groups will receive 8 weeks of two supervised therapy sessions and four home exercises per week in addition to the rehabilitation programme. At baseline (week 1) and after the intervention (week 10), movement control of the spine will be assessed using a tracking task and clinical movement control test battery. Questionnaires on pain, disability, fear avoidance and quality of life will be taken at baseline, after intervention and at 6- and 12 months follow-up. Repeated measures ANOVAs will be used to evaluate if a significant Group x Time interaction effect exists for the movement control evaluations.

Discussion: Sensor-based spinal controlled exergames are a novel way to train spinal movement control using meaningful and engaging feedback. The results of this study will inform clinicians and researchers on the efficacy of movement control training for patients with low back pain.

Ethics and dissemination: Ethical approval for this study protocol was obtained from the METC Brabant (protocol number NL76811.028.21).

Trial registration: Open Science Framework Registries (https://osf.io/v3mw9/), registration number: 10.17605/OSF.IO/V3MW9, registered on 1 September 2021.

KEYWORDS

low back pain, movement control, Static motor control, spine, exergaming, rehabilitation

Introduction

Low back pain is extremely common. More than 50% of the population will experience one or more episodes of low back pain during their lifetime (1). Most episodes of low back pain resolve within 6 weeks, but in some cases the symptoms return regularly. In those cases where symptoms persist for more than 3 months, there is a chronic condition with a variable course (2, 3). According to current evidence, the best treatment for low back pain is exercise, preferably in combination with education, but thus far, to the best of our knowledge there is no evidence that certain exercises work better than others (4).

In a considerable part of the intervention studies with a focus on physical exercises, "core stability" interventions are offered (5). During these interventions, patients are taught to selectively contract the deep trunk muscles (m. Transversus Abdominis and mm. Multifidi) in various postures and during various movements of the extremities. The lumbar spine is fixated in neutral lordosis in most of these exercises (6). We will refer to this type of exercise as "static motor control exercise" henceforth. Although these exercises have been shown to be effective in pain reduction, they are not superior to other physical exercise interventions (6).

Some patients with low back pain fixate their spine (i.e., they demonstrate reduced range of motion) during every-day movements (7–11). This behavior could be stimulated further with static motor control exercises. Moreover, several low back pain patients do experience problems with spinal movement control (12), i.e., adapting the direction, speed, and amplitude of spinal movement to the demands of the task at hand.

Designing exercises to improve movement control of the spine is a challenge. In a recent paper by Hooker et al., patients with low back pain received patient specific training to modify their altered movement pattern during functional activities (13). This resulted in a more normal distribution of hip, knee and spinal movements when picking up an object at shank height. Although this study shows that training can improve the relative contribution of joint movements during functional tasks in low back pain patients, it is no direct evidence that spinal movement control has improved.

Movement control over less centrally located joints, such as the elbow or knee, can be trained using functional tasks, like bringing a spoon to the mouth or kicking a ball toward a pylon. The success of the execution (not spilling the soup or knocking over the pylon) can be used as an indication of good control over the movement of the joint. Providing meaningful feedback on spinal movements is more complicated. Sensors that measure spinal movements can offer a solution (14). There are several sensor-based training systems available on the market, but currently only a few randomized controlled trials incorporating these technologies have been published (15, 16). These sensor-based training systems can be used to offer accurate real-time feedback on spinal movements, which could help to improve spinal movement control. These systems provide the possibility to train spinal movement control relatively independent without the need of intensive supervision and/or a highly experienced therapist (17). Moreover, the training sessions are relatively easy to standardize and the progression from simple toward complex movements can easily be adapted to each patient's abilities and needs. The sensor-based exergames could be more engaging and motivating than conventional motor controlexercises, which might increase therapy adherence (18).

This paper describes the protocol for a randomized controlled trial to evaluate if a sensor-based movement control intervention enhances movement control of the spine in low back pain patients to a greater extent than a standard static motor control intervention. We will assess movement control using a custom made spinal movement controlled tracking task and a clinical test battery by Luomajoki et al. (19, 20). We hypothesize that a sensor-based movement control intervention will enhance movement control of the spine in low back pain patients measured by spinal movement controlled tracking tasks to a greater extent than a standard static motor control intervention. The majority of the tests in the clinical test battery by Luomajoki et al. (19, 20) involve no spinal movement, hence we hypothesize that the static motor control group will improve more on this outcome than the sensor-based movement control group.

To confer clinical benefit beside movement control of the spine, we will also evaluate differences between the offered interventions in terms of therapy adherence, their respective effects on disability, pain intensity, fear avoidance beliefs, and health related quality of life.

Methods

Design

In this single-center quasi-randomized controlled trial, 60 low back pain patients will be quasi-randomly assigned to either the Sensor-Based Movement Control group ($n = 30$) or the Static Motor Control group ($n = 30$) (Figure 1). Both interventions are nested within a 12-week multidisciplinary rehabilitation programme for low back pain at the Military Rehabilitation Centre "Aardenburg" (MRC), Doorn, The Netherlands. This study protocol was approved by the METC Brabant (protocol number NL76811.028.21). Informed consent will be obtained from all patients prior to entry into the study by one of the investigators (BM, LV, MP). This trial received funding of the Stichting Ziektekostenverzekering Krijgsmacht (SZVK) in the Netherlands. This study design follows the recommendations of SPIRIT 2013 (Supplementary material 1).

	STUDY PERIOD					
	Enrolment	Allocation	Post-allocation			Close-out
TIMEPOINT*	*-T2*	*-T1*	*T0*	*T1*	*T2*	*T3*
ENROLMENT:						
Eligibility screen	X					
Informed consent		X				
Allocation		X				
INTERVENTIONS:						
Movement Control						
Static Motor Control						
ASSESSMENTS:						
Patient characteristics			X			
Movement control			X	X		
Questionnaires			X	X	X	X

FIGURE 1
Flowchart for participants of this study. *-T2 two weeks for T0; -T1 one week for T0; T0 week 1; T1 week 10; T2 week 26; T3 week 52.

Study setting

Patients will be recruited from both the inpatient and outpatient population of the MRC. Approximately 100 low back pain patients are treated in the Centre each year (21). With an inclusion rate of 80% and a dropout rate of 20%, inclusion can be completed in ~1 year and the final follow-up can be completed ~2 years after the start of the study. Enrolment started on 17 May 2021 and is ongoing. Data collection is in progress.

Patient and public involvement

There has been no patient and public involvement as co-producers of this study.

Randomization, blinding and treatment allocation

This study is quasi-randomized and non-concealed. Patients will be screened in the first week of the multidisciplinary rehabilitation programme (2 weeks prior to baseline data collection). Enrolment and allocation will be 1 week prior to baseline data collection after consent of the patient. Patients will be enrolled and allocated to each intervention by one of the investigators (BM, LV, MP) based on the starting date of the multidisciplinary rehabilitation programme. The research team has no influence on the starting date of each patient. If two patients start on the same date, allocation order will be alphabetically (based on the patients last name). The first five patients were allocated to the static motor control group, followed by five patients in the sensor-based movement control group and so forth until 30 patients have completed each intervention. This random allocation sequence was chosen by the investigators to keep group therapy planning feasible. Patients and therapists will not be blinded as this is practically impossible, however they will not be informed about the hypotheses of the study. The primary study outcome (spinal movement tracking error) will be calculated using a computer algorithm (custom made in D-flow, Motek, Amsterdam, The Netherlands) that will work independent of treatment allocation. Investigators involved in the statistical analysis will not be blinded to group. The clinical movement control battery tests will be recorded on video and scored by two examiners that are blinded to time (before/after intervention) and allocation.

Participants

The in- and exclusion criteria of this study are presented in Table 1. Patients will be screened by a physician and a manual therapist at the MRC. Based on history, physical examination and evaluation of at least one medical image obtained in the past 12 months (X-ray, CT, MRI), serious pathology of the spine will be excluded. A high Body Mass Index (BMI) could hamper the planned sensor-based movement control intervention as a result of movement artifacts (22); therefore, patients are excluded if the BMI is higher than 35 (kg/m^2). To avoid the risk of electromagnetic interference with the inertial sensors, patients

TABLE 1 In- and exclusion criteria.

Inclusion criteria
Between 20 and 60 years of age
Experienced low back pain on a daily basis over the last 3 months, with or without accompanying leg pain above the knee
Exclusion criteria
Any condition (other than chronic low back pain) that might interfere with motor control of the spine
A recent (<5 years) surgical intervention of the spinal column or a spinal fusion.
Proven serious pathology of the spine and related structures, infections, recent fractures
Psychiatric disorders
Signs of neurological compression; loss of sensory or motor functions in the legs and/or pelvis and/or radiating pain in the lower leg and/or foot
The use of drugs that influence the reaction time
A body mass index of 35 (kg/m^2) or more
Implanted electronic devices of any kind

with implanted electronic devices of any kind are excluded from this study (23).

Sample size

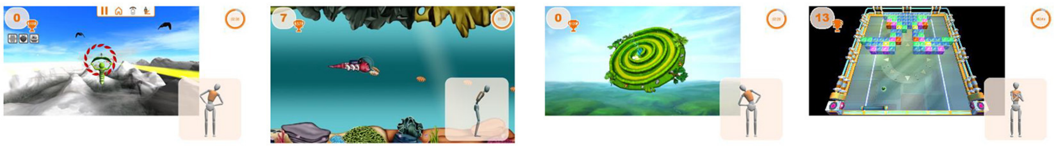
The main objective of this study is to assess if a Group x Time interaction effect exists for movement control of the spine, i.e., if movement control of the spine changes differently between groups over the course of the intervention. To the best of our knowledge, currently no studies have been performed in which spinal movement control, as defined in the current paper, is both trained and assessed before and after training, which complicates the estimation of an expected effect size. For the study outcome to be clinically meaningful we have set the goal to detect or reject an arbitrary effect size of 0.25 (24). In other words, if the effect size would be below 0.25, we would consider this result to be too small to be of interest. Differences in movement control of the spine in low back pain patients and healthy controls in terms of tracking error of a spinal movement controlled tracking task were reported by Willigenburg et al. (12). The tracking error in healthy controls was 0.332 degrees (SD 0.103) and in low back pain patients 0.422 degrees (SD 0.634), which is a large effect size (>0.8). No data about the expected effect of a sensor-based movement control intervention on these outcomes are available. However, a recent study from Matheve et al. demonstrated that low back pain patients can alter their movement behavior using a sensor-based intervention during a single session (14). Hence, we expect that low back pain patients will be able to perform equally well on movement control tasks by the end of the intervention as healthy controls without an intervention. If the static motor control group reaches 75% of the effect of the sensor-based movement control group, corresponding to an effect size of 0.25 (considering the effect of the intervention is equally large as the standard deviation of the effect), a total sample size of 54 (27 per group) would suffice to demonstrate a Group x Time interaction effect at a power of 95%. In case of a drop-out an additional patient will be recruited (with a maximum of 10 patients).

Interventions

The Sensor-Based Movement Control and Static Motor Control intervention will be offered over a course of 8 weeks (weeks 2–9 of the study), each week consisting of two supervised therapy sessions of 20–30 min and four non-supervised home exercises of 5–10 min resulting in nearly one training session every day of the week for 8 weeks.

The supervised sessions are provided by five experienced (4–10 years) physio- and occupational therapists from the MRC that are trained to provide both the intended interventions. The therapists can also provide the regular therapies in the multidisciplinary rehabilitation programme of a patient. Three training moments are given to all therapists at the same time by the investigators to understand the content of the interventions and how to offer this to the patients during 8 weeks. The quality will be assured by several evaluation meetings with the investigators and investigators will check the content of the sessions by occasionally being present at the supervised sessions throughout the study. The first four sessions will be individual, i.e., one patient supervised by one therapist. In these sessions, the capacity of the patient will be determined by the therapists and the patient will get acquainted with the basics of the training. The final 12 sessions will be in groups, with a maximum of three patients in the same intervention per session, supervised by one therapist. The patients will not have the same therapist throughout the programme to keep the rehabilitation planning feasible. The therapists will monitor the progress and challenge the patient if needed during all sessions. There is a standard protocol for exercises throughout the sessions, for both the sensor-based movement control and static motor control group. This protocol was composed by two experienced physiotherapists/human movement scientists and are based on literature (6). Therapists are allowed to modify the standard exercises to match the difficulty level to the capacity of the patient. The standard exercises can also be deviated from to the need of the individual patient as long as it is within the scope of the assigned intervention. There is no standardized approach for modifications in progression, however all modifications will be registered. An example of the protocol for week four is presented in Figure 2. Attendance of the supervised sessions will be registered.

	Exercise 1	Exercise 2	Exercise 3	Exercise 4
Movement control	**Glider** Trunk lateral flexion in standing position for 3 minutes.	**Diver** Trunk flexion/extension in standing position for 3 minutes.	**Maze** Flexion/extension + lateral flexion in standing position for 3 minutes.	**Brick Breaker** Trunk rotation in standing position for 3 minutes.



Static motor control	**Bridge exercise** 3 times 15 repetitions.	**Superman exercise** 3 times 5 repetitions per arm and per leg.	**Planking exercise** Maintain position 3 times, for 5 seconds with 10 repetitions.	**Dynamic knee raise** 3 times 10 repetitions per leg.

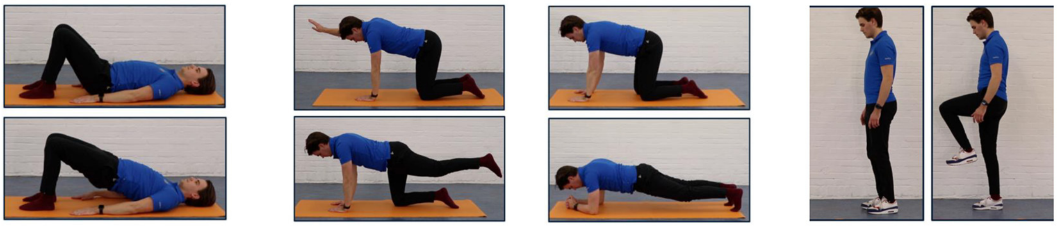


FIGURE 2
Supervised exercises offered halfway through the intervention (week 4). The exercises of the sensor-based movement control group are displayed in the top row, the exercises of the static motor control group in the bottom row.

The home exercises consist of four exercises per week, that will be performed on non-therapy days. The non-supervised home exercises are listed in a workbook with text and pictures describing each exercise. In addition, a QR-code is added to each exercise linking to a video-instruction. These video-instructions will be available "unlisted" on YouTube. The therapists will encourage patients in their own manner to do their home exercises. Adherence to the home exercises will be measured by a questionnaire after the intervention and at follow-up at 26 and 52 weeks.

The full protocol of supervised sessions and home exercises is provided in Supplementary materials 2, 3.

Static motor control intervention

The supervised therapy sessions and home exercises of the standard static motor control intervention will consist of exercises in which patients will be instructed to contract their m. Transversus Abdominis during a variety of postures and body movements while keeping their spine in neutral position, i.e., trying to make as little spinal movements as possible. The exercises will be offered with increasing intensity, difficulty, and complexity per week by using a variety of postures, movements, and exercise equipment such as a balance board or foam pad to stand on. Modifications in progression of the patient will be registered by the treating therapist. The home exercises will be covered during the supervised sessions to adjust the load level of these exercises to the capacity of the patient if necessary.

Sensor-based movement control intervention

For the supervised therapy sessions of the Sensor-Based Movement Control intervention, Valedo® Motion 2.0 (Hocoma) will be used. Valedo Motion is a medical device on which a patient can play games controlled with spinal movements. Spinal movements are tracked using three small inertial measurement units (IMUs), placed on the pelvis (S1), thorax (sternum) and thoracolumbar (L1) area of the spine. The orientation of these sensors is streamed to a laptop and used in real-time to control several games. These games are displayed on a laptop and invite the player to make controlled (in terms of movement direction, speed, and amplitude) movements of the spine in various postures (e.g., standing, sitting or on hands and knees). During the first few sessions, the workflow of the hardware and software will be demonstrated and explained by the therapist. It is expected that the patient can perform

TABLE 2 Overview of outcome measurements in this study.

	Intervention		Follow-up	
	T0	T1	T2	T3
Patient characteristics	X			
Age, height, weight, BMI, gender, duration of complaints				
Questionnaires	X	X	X	X
RAND-36, FABQ, NRS, ODQ, RMDQ				
Movement control assessment	X	X		
Tracking tasks, clinical spinal movement control tasks, repetitive motion tasks and gait trials				
Therapy adherence		X		
EARS				

T0: week 1; T1: week 10; T2: week 26; T3: week 52.

the set-up independently (under supervision of the therapist) after these two sessions. During each session, patients will play four different games. Before each session, the patients spinal range of motion around the three anatomical axes will be determined using the software. The standard protocol for games throughout the sessions are pre-set by the investigators and will be offered with increasing intensity, difficulty, and complexity. Modifications in progression of the patient will be registered by the treating therapist. The home exercises resemble the movements and postures of the Valedo games of that week. These exercises will also be adjusted to the capacity and need of the patient by the therapists, for example by changing the game duration or difficulty level of the game or by using exercise equipment.

Multidisciplinary rehabilitation programme

All patients of the study will be enrolled in a multidisciplinary rehabilitation programme. This is a standard care programme at the MRC for patients with chronic low back pain which has a focus on increasing the activity and physical level, education about back pain, healthy lifestyle and awareness of the body and physical limits. The programme follows a protocol in which the number, duration and content of the therapies is fixed. During this 12-week programme, they will receive multiple therapy sessions for 3 days a week. The programme mainly consists of physiotherapy (19 individual sessions of 30 min), occupational therapy (19 individual sessions of 30 min), sports therapy (20 group sessions of 60 min consisting of fitness, swimming and game sports) and 4 group sessions of body awareness. In the *first* week, pain-education is given by a psychologist and social worker and, if necessary, further individual guidance is provided once a week. The therapists of these disciplines are discouraged, but not prohibited, to focus their interventions on static motor control or spinal movement control exercises and they will not have access to Valedo® Motion during these therapy sessions. Moreover, they are request not to compensate for the given intervention (e.g., providing more dynamic exercises for patients in the static movement control group). The therapists will not be restricted in their therapy programme in any other way. It will not be registered to what extent static motor control or movement control exercises are provided during these therapy sessions.

Data collection and outcome measures

Patients will be tested at two instances, once before (T0: week 1) and once after the intervention (T1: week 10). In addition, we will contact them by email at 6- and 12-months follow-up (T2: week 26 & T3: week 52). Per follow-up moment, patients receive a maximum of 2 emails and 1 letter, to enhance completion of the follow-up. Table 2 highlights the measures collected at each point in time. At the start of the intervention study, patients' characteristics will be recorded to enable comparison of baseline characteristics of both groups.

Primary outcome

Our primary outcome measure is movement control of the spine. This will be quantified using three spinal movement controlled tracking tasks and a clinical movement control test battery. The movement controlled tracking tasks are based on the tracking task used in the study of Willigenburg (12). The tracking tasks used in this study consist of one flexion-extension, one lateral flexion and one rotation task, and will be performed at T0 and T1 with 3 Valedo Motion inertial measurement units (IMUs) attached at the right thigh, pelvis (at S1 level) and at the sternum level. During these tasks, patients will be instructed to move their spine in order to keep its real-time representation (on a laptop screen located in approximately one meter in front of them at eye level) within a moving target. Patients are in a seated position. The patient's pelvis will be fixated with a frame, which will be used to guarantee that the spinal angle is changed without any hip motion. Each trial will last 2 min and 40 s, with the first 40 s being for learning the task, and the following 2 min for the actual measurement. During the flexion/extension task, the vertical position of the target on the screen will vary between values that correspond to 20 degrees trunk flexion and 10 degrees trunk extension. During the lateral flexion and rotation task, the horizontal position of the target on the screen will vary between values that correspond to 10 degrees left and 10 degrees right lateral flexion or rotation. In each tracking task,

TABLE 3 Used offset and ROM for each movement plane and characteristics of the multi-sine-wave.

Movement plane	**Off**set	**ROM**	Excursion
Sagittal (flexion)	5	30	−10/20
Frontal (lateral flexion)	0	20	−10/10
Transversal (rotation)	0	20	−10/10
Sine no.	***Ampl***itude (%)	***f***requency (Hz)	***φ***: Phase (rad)
'Main' Sine #1	80	0.025	0.00
Sine #2	10	0.215	0.22
Sine #3	6	0.185	0.14
Sine #4	4	0.250	0.84

Bold text in the table corresponds to bold text in Formula 1. ROM: Range of Motion.

the target will follow a multi-sine with a main frequency of 0.025 Hz (one cycle each 40 s). All these movement excursions are within the maximum range of the tracking task that was used by Willigenburg (12). Formula 1 and Table 3 describe the movement profile of the tracking target in each task, illustrated in Figure 3. The reliability or minimal detectable changes of this movement control measurement is unknown.

Formula 1: Used offset and ROM for each movement direction and characteristics of the multi-sine-wave. *Bold text corresponds to bold text in Table 3. t = time, starting at the beginning of the tracking task.*

$$Target_{plane}(t) = \mathbf{Off}_{plane} + \frac{\mathbf{ROM}_{plane}}{2} \times \sum_{i_{Sine}=1}^{4} \mathbf{Ampl}\,(i_{Sine}) \times sin\,(2\pi\,\mathbf{f}\,(i_{Sine})\,t + \boldsymbol{\varphi}\,(i_{Sine}))$$

The tracking error (average absolute deviation from imposed trunk angle) around the imposed movement axis of the three tasks will be reported as average tracking error (in degrees).

The clinical movement control test battery of the lower back will be the tests of Luomajoki et al., (19, 20). This test battery consists of six active movement control tests in which the patient performs each movement once. The test will be recorded on video from the front or side (depending on the movement) using a video-recorder and two blinded experienced (>10 years) physiotherapists will rate the performance of the tests from the recordings. These therapists will be instructed by the investigators how to score this assessment prior to the ratings. The total score can range from 0 to 6, indicating the number of tests with clear movement dysfunction. The final score of the test battery will be calculated as a mean of the two raters.

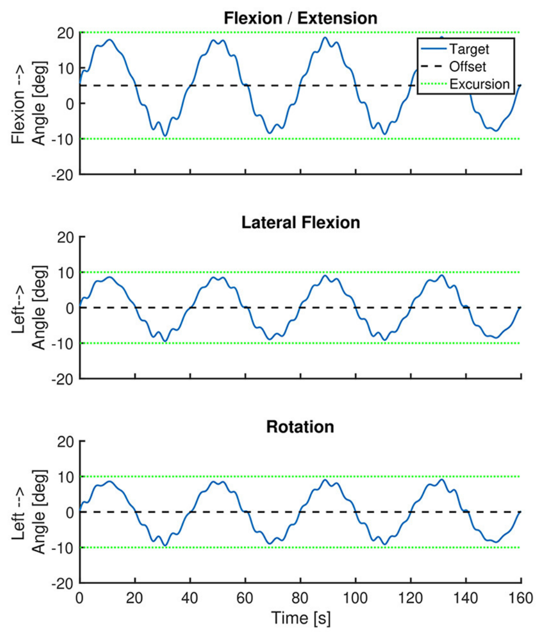


FIGURE 3
Position of the tracking target over time. The blue line represents the position of the tracking target over time. The black dashed line represents the offset of the target trajectory. The green dotted lines represent the movement excursion during each task. The imposed movement excursion during the lateral flexion and rotation task was symmetrical. During the flexion/extension task the movement excursion was larger in the flexion direction (i.e., 20 degrees) than in extension direction (i.e., 10 degrees).

Secondary outcomes

Spinal movement will also be measured in other movement tasks as secondary outcomes.

- Gait trials: Cycle-to-cycle variability of spinal rotations (measured in degrees) will be measured during gait on a treadmill. The tasks consist of walking for 5 min at three different walking speeds, with a small pause for IMUs calibration in between: one at comfortable walking speed, one at 6 km/h ("fast walking") and one at 2 km/h ("slow walking") (25).
- Repetitive bending task: Cycle-to-cycle variability of spinal flexion (measured in degrees) will be measured during a repetitive bending task. Patients will touch the sides of a box (2 cm x 30 cm x 30 cm) that is in front of them, at their tibial tuberosity height, for 40 times. Prior to the task, a video of the task executed at 0.92 Hz is shown to give patients a visual demonstration of the expected task movement and frequency.
- Repetitive rotation task: Cycle-to-cycle variability of spinal rotation will be measured during a repetitive standing rotation task. Patients will touch two lateral targets with

their contralateral hands, alternating them for 80 times (40 times each). Targets' positions are at the patients' homolateral arm distance, at their shoulders' height, and rotated 45° relative to their anteroposterior axis. A video of the task executed at 0.62 Hz is shown to give patients a visual idea of the task's expected movement and frequency.

– Self-developed movement control test of the lower back. During this test, patients will perform 4 movements consisting of 3 continuous repetitions of a pelvic tilt, flexion/extension, lateral flexion and rotation of the spine. First, all 4 movements will be performed seated and next, these same 4 movements will be performed in standing position resulting in 8 tests. Each test will be scored in "correct" (two points): low back or pelvic movement is performed fluently and isolated (thoracic movement in absence of pelvic movement or vice versa); "partial correct" (one point): the movement is performed not fluently *or* insufficiently isolated; or "not correct" (zero points): the movement is not performed fluently nor isolated. A higher score represents a better movement control of the lumbar spine. The test will be recorded on video and rated in the same manner as the movement control test battery of Luomajoki. This test was developed because in the test battery by Luomajoki et al. (19, 20) no movement of the spine is requested during most tests. In fact, in five out of the six imposed movements, the subjects are instructed explicitly to not move the lumbar spine.

We have no specific hypotheses regarding the aforementioned movement variability outcomes. These outcomes were primarily assessed as part of a case control study (see pre-registration https://osf.io/3dr58).

The tracking tasks, gait trials and repetitive motion tasks will be performed in quasi-random order at baseline and post-intervention measurements.

In addition to the movement control outcomes, patient reported outcome questionnaires will be assessed at baseline, post-intervention and at follow-up. The patient reported outcomes measured in this study will be: Exercise Adherence Rating Scale (EARS) (26), Dutch version of the Oswestry Disability Index (ODI) 2.1a (27), Dutch version of the Roland Morris Disability Questionnaire (RMDQ) (28, 29), three scores of Numeric Rating Scale (NRS) for Pain; the average and maximum pain intensity over the past 7 days and current pain intensity (29), Dutch version of Fear Avoidance Beliefs Questionnaire (FABQ) (30), Four scales of the Dutch version of the RAND-36 (Physical functioning, mental health, general health and pain) (31).

Data management

Patients will receive a unique three-digit number that will be used on all forms (except the informed consent form) used in this study. Only the principal investigators will have access to the key of this code list. The informed consent and patient related forms will be stored separately from the other forms and will be stored for 15 years. Video-recordings of the clinical movement control test battery will be stored locally at the MRC. Only the principal investigators will have access to these recordings. The recordings will be scored by two independent physical therapists, under supervision of a principal investigator.

Statistical analysis

To evaluate if movement control of the spine changes differently between groups over the course of the treatment, repeated measures ANOVAs will be used to evaluate if a significant Group x Time interaction effect exists for the tracking error during each tracking task separately, the average tracking error of the tracking tasks and the total score of the clinical movement control test battery. The main effects of Group and Time will also be assessed using the same ANOVA. In addition to the total score of the clinical movement control test battery by Luomajoki et al. (19, 20) the performance on the individual tests that comprise the test battery will also be reported per group at T0 and T1.

Secondary study parameters will be assessed in the same manner as described above without correction for multiple testing, because these analyses are of an exploratory nature, and we want to limit the probability of type 2 errors. For the questionnaires that will be filled out on more than two occasions, we will perform *post-hoc* independent t-tests, with LSD correction, comparing results between groups at each point in time. Patient characteristics and all questionnaires filled out during the first testing day will be compared between groups using independent sample *t*-tests without correction for multiple testing to evaluate if differences existed at baseline. Statistical analyses in this study will not be adjusted for baseline differences between groups, as recommended by de Boer et al. (32). Independent of normality, parametric statistics will be used in this study and the alpha level will be set at 0.05 (33).

Data from patients who attended less than 10 sessions are not included in the statistical analysis. In addition, the data is also not included if patients have dropped out of the study before T1.

Missing data will be handled by using complete case analysis with all repeated measures ANOVAs between T0 and T1 and independent *t*-tests between T0, T1 and T2 and between To, T1, T2 and T3.

All statistical analyses will be performed using R version 4.1.1.

Adverse events

Adverse events are defined as any undesirable experience (harmful, objectionable, or unpleasant) occurring to a patient during the study, whether or not considered related to the testing procedures or the experimental intervention. In the study information letter patients are instructed to contact the investigators in case of an adverse event. All adverse events reported spontaneously by the patient or observed by the research team will be recorded. A serious adverse event is any untoward medical occurrence or effect that results in death; is life threatening (at the time of the event); requires hospitalization or prolongation of existing inpatients' hospitalization; results in persistent or significant disability or incapacity; any other important medical event. The investigator will report all SAEs to the sponsor without undue delay after obtaining knowledge of the events.

Data monitoring

The study will be terminated prematurely if decided so by: the board of physiatrists from the MRC or the board of the MRC. There is no data monitoring committee or independent audit for this study.

Discussion

In this study, the effect of a sensor-based spinal movement control intervention on the movement control of the spine in low back pain patients over the course of a multidisciplinary rehabilitation programme will be compared to the effect of conventional static motor control exercises. In addition, we aim to evaluate the effect of the intervention on disability, pain intensity, fear avoidance beliefs and health related quality of life. Finally, therapy adherence will be compared between the interventions. Sensor-based exergames are a relatively new tool to train spinal movement control using meaningful and engaging feedback. To our acknowledge, this is the first study which evaluates if sensor-based exergames training influences movement control of the spine in low back pain patients to a greater extent than static motor control training.

Currently, there is no gold standard to assess movement control of the spine. Therefore, we will analyse our main outcome with three different assessments: [1] sensor-based tracking tasks on a laptop, based on a tracking task from Willigenburg et al. (12), [2] the clinical movement control test battery of Luomajoki et al. (19, 20) and [3] a self-developed clinical movement test battery. Because these tests are performed in the same subjects at the same moment, the results of this study could provide us more insight in how to assess movement control of the spine. The reliability and minimal detectable changes of the movement control tracking tasks and the self-developed clinical movement test battery are not available which may bias the outcome of the study.

There are several limitations of this study that need to be addressed. Our study population completely consists of Dutch military personnel. The Dutch military population mostly consist of males, who are relatively young and physically active compared to the civilian population (34). The cause of low back pain in this population is mostly overuse, due to the high workload in the Netherlands Armed Forces (34, 35). For this reason, the generalizability of the results of this study might be compromised. Another limitation, from a clinical perspective, is that the main outcome of this study (spinal movement control) does not correspond to the main focus of most patients, which is reducing pain and/or disability. Although these outcomes will be assessed, the study might be underpowered to demonstrate significant effects on these domains for at least two reasons. First, some patient subgroups may derive a greater benefit from one type of exercise than another (e.g., static motor control vs. sensor based movement control) because of the heterogeneity in the low back pain population. Exploratory analyses can be performed to evaluate if these subgroups appear to exist, but a larger sample is expected to be required to provide more conclusive evidence. Second, because this study is embedded in a multidisciplinary rehabilitation program, an effect of the tested intervention on pain or disability might be hidden by the effect of other components of the program. Finally, for the main outcome of this study, movement control of the spine, the effect of other therapies might reduce the contrast between groups as well. It cannot be excluded that patients in the static motor control group will also receive exercises of the movement control group during other therapies and vice versa. This can be considered a study confounder. However, we hypothesize that it is relatively difficult to improve spinal movement control without the use of sensors. Embedding this study within a multidisciplinary rehabilitation program could be considered a test of this assumption. Each week of the intervention consists of two supervised therapy sessions of 20–30 min and four non-supervised home exercises of 5–10 min. Although therapists will encourage patients to do their home exercises, compliance can influence results in this study. For this reason, self-reported exercise adherence will be measured in both groups.

The results of this study will help to inform clinicians and researchers on the efficacy of movement control training in combination with multidisciplinary rehabilitation programme for patients with low back pain. Also it will enlighten preliminary impacts of the interventions on patient reported outcomes. This could directly affect decision making in clinical practice and culminate in larger trials to assess if pain and/or disability could be reduced by movement control training in (subgroups of) low back pain patients.

Ethics and dissemination

Ethical considerations

This study will be performed in accordance with the Declaration of Helsinki. Ethical approval was obtained from the Medisch Ethische Toetsingscommissie (METC) Brabant on 14 Mai 2021 and all procedures will be conducted in accordance with the statement conducting research involving humans. Informed consent will be obtained by the investigators from all potential patients and patients will be aware that participation is voluntary and can withdraw from the study at any time.

Safety considerations

The tests at the beginning and end of the intervention could result in a transient increase in low back pain. Training with sensors could result in spinal tissue overload as a result of lack of focus on bodily sensations. However, the Military Rehabilitation Center has more than 10 years of experience with providing a similar type of therapy in low back pain patients. Moreover, the complexity, duration and intensity of the exercises will be increased gradually, which would minimize the chance of overloading the spinal structures. Damage to research subjects through injury or death caused by the study is covered by the Ministry of Defense. This applies to the damage that becomes apparent during the study or within 4 years after the end of the study.

Dissemination

To protect confidentiality, personal information about the patients will be collected, shared and maintained in a database on a secured computer that can only be accessed by principal investigators before, during and after the trial.

Any significant modifications of the study protocol will be communicated to the METC, trial funder (SZVK), Open Science Framework Registries and the trial sponsor (MRC). The investigators will communicate trial results to the patients, trail sponsor, METC and funder within 1 year after the end of the study. The study results shall be presented at symposia, conferences and to publish in journals and theses without publication obligations from the sponsor.

Ethics statement

The studies involving human participants were reviewed and approved by Medisch Ethische ToetsingsCommissie (METC) Brabant. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

Author contributions

BM, LV, SB, JD, and MP contributed to conception and design of the study. BM and LV are the coordinating researchers and responsible for recruitment, randomization, data collection, and writing the first draft of the manuscript. MP supervises the study and also drafts sections of the manuscript. All authors contributed to manuscript revision, read, and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2022.1010054/full#supplementary-material

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Biomechanics and lower limb function are altered in athletes and runners with achilles tendinopathy compared with healthy controls: A systematic review

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Achilles tendinopathy (AT) is a debilitating injury in athletes, especially for those engaged in repetitive stretch-shortening cycle activities. Clinical risk factors are numerous, but it has been suggested that altered biomechanics might be associated with AT. No systematic review has been conducted investigating these biomechanical alterations in specifically athletic populations. Therefore, the aim of this systematic review was to compare the lower-limb biomechanics of athletes with AT to athletically matched asymptomatic controls. Databases were searched for relevant studies investigating biomechanics during gait activities and other motor tasks such as hopping, isolated strength tasks, and reflex responses. Inclusion criteria for studies were an AT diagnosis in at least one group, cross-sectional or prospective data, at least one outcome comparing biomechanical data between an AT and healthy group, and athletic populations. Studies were excluded if patients had Achilles tendon rupture/surgery, participants reported injuries other than AT, and when only within-subject data was available.. Effect sizes (Cohen's *d*) with 95% confidence intervals were calculated for relevant outcomes. The initial search yielded 4,442 studies. After screening, twenty studies (775 total participants) were synthesised, reporting on a wide range of biomechanical outcomes. Females were under-represented and patients in the AT group were three years older on average. Biomechanical alterations were identified in some studies during running, hopping, jumping, strength tasks and reflex activity. Equally, several biomechanical variables studied were not associated with AT in included studies, indicating a conflicting picture. Kinematics in AT patients appeared to be altered in the lower limb, potentially indicating a pattern of "medial collapse". Muscular activity of the calf and hips was different between groups, whereby AT patients exhibited greater calf electromyographic amplitudes despite lower plantar flexor strength. Overall, dynamic maximal strength of the plantar flexors, and isometric strength of the hips might be reduced in the AT group. This systematic review reports on several biomechanical alterations in athletes with AT. With further research, these factors could potentially form treatment targets for clinicians, although clinical approaches should take other contributing health factors into account. The studies included were of low quality, and currently no solid conclusions can be drawn.

KEYWORDS

achilles tendinopathy, biomechanics, neuromuscular, kinetics, electromyography, athletes, runners, kinematics

1. Introduction

Achilles tendinopathy (AT) is a debilitating overuse injury, symptoms of which can include pain localized to the Achilles tendon, morning stiffness, and functional impairments during dynamic activities such as running and hopping (1, 2). Achilles tendinopathy is a recurrent problem for both athletic and non-athletic populations (3–5). Whilst a study by Lysholm et al. (6), reported a 9% annual incidence of Achilles disorders in runners, a different investigation found an AT point prevalence of 36% in approximately 1,000 runners (7). A separate study by Albers et al. (4), found that 65% of AT cases do not involve sport. Current research therefore indicates that mechanisms of AT development might be multi-factorial in nature and injury presentation may differ according to population category i.e., athletic vs. non-athletic (3).

Clinical risk factors for AT have been discussed in a recent publication by van der Vlist et al. (5), and include prior lower limb tendinopathy or fracture, use of ofloxacin antibiotics, moderate alcohol consumption, increased time between heart transplantation and initiation of treatment for infectious disease, as well as cold weather training. Furthermore, various neuromechanical indications relating to human biomechanics seem to increase the risk of AT. These neuromechanical factors may be manifested in decreased isokinetic plantar flexor strength, and abnormal gait pattern with decreased forward progression of propulsion and more lateral foot-roll over at the forefoot flat phase (5). Such factors may be of particular relevance for athletic populations, especially for those engaged in activities that require repetitive stretch-shortening-cycle loading (SSC), such as running and jumping (2, 8, 9). It has been hypothesized that repetitive loading of the Achilles tendon, which is not compensated *via* sufficient strength or endurance of the plantar flexor muscles or optimal gait biomechanics, may result in injury (1, 2, 5, 9–11). This has led to the widespread implementation of biomechanically-driven and strength-based loading programs in the rehabilitation and prevention of AT (12–16). However, it is important to understand this model in the context of other potential contributing factors to AT overuse injury, such as increasing age (17), training load (18), increased BMI (19) and other considerations mentioned previously (5).

Several systematic reviews and meta-analyses have recently investigated the relationship between biomechanical factors and AT (2, 10, 20, 21). In two independent meta-analyses, Hasani et al. (21), and McAuliffe et al. (10), concluded plantar-flexor strength deficits to be associated with AT, when compared within-subject (affected vs. healthy limb) or with healthy controls. Although, deficits were more pronounced between sides than when compared with the control group in the more recent analysis (21). Two further systematic reviews focused on aspects of gait and lower-limb biomechanics (2, 20). Sancho et al. (2019) (2), reported biomechanical alterations in AT patients during running and hopping after conducting a meta-analysis across 16 studies, including changes in kinematics, kinetics and muscle activity. A similar review including 14 studies found comparable results regarding alterations in gait in AT patients (20). It should be noted that both reviews indicated a high risk of bias across studies and recognized a lack of high-quality prospective research in the area. Another interesting avenue of enquiry is adaptations of reflex responses in patients with AT, and two prominent studies have suggested higher volitional supraspinal reflexes (22) and altered central nervous system reflex regulation in tendinopathic tendons (23). Considering altered reflex responses have been observed in other persistent musculoskeletal pain disorders (24), their relevance for AT patients may warrant further exploration and review.

As described, a range of data summarised in multiple studies has revealed weak to moderate evidence that the biomechanics of patients with AT are potentially altered (2, 10, 20, 21). However, these reviews have tended to focus on a single component of human movement e.g., isolated joint strength (10, 21) or gait mechanics (2, 20), providing a useful but arguably narrower picture of the data. Thus, synthesising the evidence into a single comprehensive review could prove helpful in furthering understanding of these alterations in AT populations. Besides this, none of the above-mentioned reviews implemented a set training load within inclusion criteria e.g., running >20 km per week or equivalent, even though clinical presentation of AT may vary between athletic and non-athletic populations (3, 5). In addition, three of the reviews included studies which compared parameters associated with AT between sides within the injured group (affected vs. healthy limb) (2, 10, 21), despite evidence suggesting that the contra-lateral healthy limb might also present with sensory motor deficits in tendinopathy patients

(25) and research indicating central sensitization and altered central pain processing in AT (26, 27). Furthermore, in two of the previously conducted reviews there was no set criteria for AT diagnosis stated within the inclusion criteria of investigated studies (10, 20), although best practice diagnosis guidelines have previously been outlined (28, 29).

Therefore, the aim of this study was to conduct a systematic review, with the goal of synthesising information regarding biomechanical alterations and changes in lower limb function in specifically athletic populations with AT, when compared to an asymptomatic, athletic, healthy control group. Populations in both groups were defined as athletic, based upon strict inclusion criteria.

2. Methods

This systematic review was conducted and reported in accordance with the PRISMA (Preferred Reporting Items for Systematic Reviews and Meta-Analyses) guidelines (30). The review was not pre-registered.

2.1. Search strategy

The electronic databases MEDLINE, Web of Science and Cochrane Library were searched in March 2021. Two authors (A.Q., J.M.) completed the initial search of all databases simultaneously, after critical discussion of key terms and development of a search strategy. The four categories identified as the base of the search strategy were "Biomechanics", "Movement Task", "Pathology (Tendinopathy)" and "Anatomical Location (Achilles tendon)". MeSH terms were also applied to enabled search terms and in databases which featured this function. Filters of (1) Human subjects/Not animals, (2) Language: Only English or German articles and (3) Research published in the last 20 years (2001–2021) were applied either directly as search terms or within filter settings of the corresponding database. Search terms for MEDLINE (PubMed) are detailed in **Table 1**.

2.2. Eligibility criteria

Study eligibility was determined based upon strict inclusion and exclusion criteria, which were defined as follows:

Inclusion Criteria

- AT diagnosis based upon established guidelines – History of localised Achilles tendon pain (mid-portion and/or insertional), and at least one of the following: pain during or after activities that load the tendon, morning stiffness, and tenderness on palpation.
- Data should be cross-sectional, prospective or baseline data from intervention studies.
- Studies comparing biomechanical features during human gait, in hopping, jumping or other functional movement

TABLE 1 MEDLINE (PubMed) search strategy.

Category	Terms
Biomechanics	("Biomechanical phenomena"[MeSH]) OR ("Biomechanic*"[Title/Abstract]) OR ("Kinematic*"[Title/Abstract]) OR ("Kinetic*"[Title/Abstract]) OR ("Kinetics"[MeSH]) OR ("Motion"[MeSH]) OR ("Temporospatial"[Title/Abstract]) OR ("Plantar pressure"[Title/Abstract]) OR ("Ground reaction force"[Title/Abstract]) OR ("GRF"[Title/Abstract]) OR ("Moment"[Title/Abstract]) OR ("Torque"[MeSH]) OR ("Force"[Title/Abstract]) OR ("Stiffness"[Title/Abstract]) OR ("3D Kinematics"[Title/Abstract]) OR ("Mechanic*"[Title/Abstract]) OR ("Mechanics"[MeSH]) OR ("Muscular"[Title/Abstract]) OR ("Neuromuscular"[Title/Abstract]) OR ("Neuro-muscular"[Title/Abstract]) OR ("Neuromotor"[Title/Abstract]) OR ("Neuromotor control"[Title/Abstract]) OR ("Motor control"[Title/Abstract]) OR ("Reflex" [MeSH]) OR ("Reflex, stretch"[MeSH]) OR ("EMG"[Title/Abstract]) OR ("Electromyograph*"[Title/Abstract]) OR ("Electromyography"[MeSH]) OR ("Muscle activit*"[Title/Abstract]) OR ("Strength*"[Title/Abstract]) OR ("Muscle strength"[MeSH]) OR ("Weak*"[Title/Abstract]) OR ("Strong*"[Title/Abstract]) OR ("Power*"[Title/Abstract]) OR ("Muscle*"[Title/Abstract]) OR ("Muscles"[MeSH]) OR ("Function*"[Title/Abstract]) OR ("Endurance"[Title/Abstract]) OR ("Fatigue*"[MeSH]) OR ("Muscle fatigue"[MeSH]) OR ("Stiff*"[Title/Abstract]) OR ("Rate of force development"[Title/Abstract]) OR ("RFD"[Title/Abstract]) OR ("Stress"[Title/Abstract]) OR ("Strain"[Title/Abstract])
Movement Task	("Running"[MeSH]) OR ("Walking"[MeSH]) OR ("Gait"[MeSH]) OR ("Running Gait"[Title/Abstract]) OR ("Gait-related" [Title/Abstract]) OR ("Gait related" [Title/Abstract]) OR ("Locomotion"[MeSH]) OR ("Bounc*"[Title/Abstract]) OR ("Plyometric*"[Title/Abstract]) OR ("Plyometric exercise"[MeSH]) OR ("Jump*"[Title/Abstract]) OR ("Hopping"[Title/Abstract]) OR ("Hop"[Title/Abstract]) OR ("Land*"[Title/Abstract]) OR ("Drop*"[Title/Abstract]) OR ("Isokinetic*"[Title/Abstract]) OR ("Concentric*"[Title/Abstract]) OR ("Eccentric*"[Title/Abstract]) OR ("Isometric*"[Title/Abstract]) OR ("Isometric contraction"[MeSH]) OR ("Resistance Training"[MeSH])
Pathology (Tendinopathy)	("Tendinopathy"[MeSH]) OR ("Achilles Tendinopathy"[Title/Abstract]) OR ("Tendinitis"[Title/Abstract]) OR ("Tendinosis"[Title/Abstract])
Anatomical Location (Achilles tendon)	("Achilles tendon"[MeSH]) OR ("Plantarflex*"[Title/Abstract])
(Human Subjects)	NOT (Animals)

The categories were combined using the Boolean command "AND".

activity, during isolated strength activities or measuring reflex activity between AT patients and healthy asymptomatic controls.

- Population should be an athletic/recreational athletic population in regular training e.g., >20 km running/ training >2 h a week. Sport should include repetitive SSC load on Achilles tendon.
- Articles in English or German.

Exclusion Criteria

- Participants with Achilles tendon rupture and/or surgical intervention.
- Studies including participants with injury other than Achilles tendinopathy.
- Reviews, case-series, case studies, opinion articles and abstracts.
- Studies comparing within-subject e.g., injured vs. non-injured leg.

2.3. Selection process

All studies were screened independently by two of the authors (A.Q., J.M.). Titles and abstracts of all obtained records were downloaded into an electronic reference management software (Mendeley Desktop 1.19.4). Duplicates were removed with the aid of the automatic detection system within the reference manager and manually checked. Titles and abstracts of studies were matched against pre-defined inclusion and exclusion criteria for eligibility. Articles included by title and abstract were then assessed for inclusion by full text, and if no reason for exclusion was discovered the articles were included for synthesis within the systematic review. Any disagreements on study inclusion or exclusion were discussed and resolved between the two authors (A.Q., J.M.) in conversations arbitrated by a third author (T.E.). The reference lists of included studies and relevant systematic reviews were also searched to look for potential studies that might meet inclusion criteria.

2.4. Risk of bias assessment

The risk of bias for included studies was assessed using the Critical Skills Appraisal Programme Case-Control Study Checklist (31). The original checklist contains 12 questions but only 10 were relevant to the studies within this systematic review. Therefore, these 10 questions were applied to assess the quality of the included studies, as adapted previously in a similar systematic review (10). A list of the questions, their associated criteria and scoring strategy are provided in **Supplemental File S1**. The risk of bias assessment was performed independently by two authors (A.Q., J.M.). Disagreements on scoring of the individual studies were managed by consensus and if agreement could not be reached, a third author (T.E.) was consulted to resolve the debate.

2.5. Data extraction and analysis

A pre-defined data extraction sheet was prepared with the following variables: sample size, participant demographics and details (e.g., age, sex, anthropometrics, training status, AT diagnosis and symptoms duration), study design, characteristics of the task investigated, biomechanical variables studied and any reported significant findings. Data were extracted by each reviewer (A.Q., J.M.) for the included studies. Study results were then sub-categorised by the task characteristics investigated, to allow for a synthesis of variables within each pre-defined area of motor behaviour. Task characteristics were categorised as (1) "Gait" – running/ walking, (2) "Non-gait multi-joint activity" – hopping/ jumping/squatting, (3) "Isolated joint strength", (4) "Reflex activity" – specific methodologies targeting reflex responses. Relevant biomechanical variables associated with the specific movement behaviour, were then reported within each category. In cases where studies reported on more than one relevant task, findings from the single study were extracted, separated, and binned into the appropriate category. In cases where data needed for inclusion was not found within the manuscript, the relevant authors were contacted to obtain the specific details required. When available, means and standard deviations were extracted from included articles. These were used to calculate effect sizes (Cohen's d) with corresponding 95% confidence intervals, allowing for better comparison between the studies (reported as: Effect Size (ES) d [95% Confidence Intervals (CIs)]). Effect sizes were considered statistically significant (indicated with*), if 95% confidence intervals did not cross the zero level.

3. Results

3.1. Identification of studies

A total of 4,442 studies were yielded with the initial search criteria. After excluding studies *via* title and abstract, 59 full-text studies were identified as potentially suitable. Twenty of these full-text reports (taken from 19 experimental study populations) met the inclusion criteria and were synthesised for data extraction and analysis within this review. Details of this process can be seen in the PRISMA flow diagram (**Figure 1**). Eighteen of the included studies were case-control study designs, whilst 2 of the studies were prospective (32, 33). Concerning task characteristics, 11 of the studies examined "Gait" characteristics (32–42), 3 studies reported on

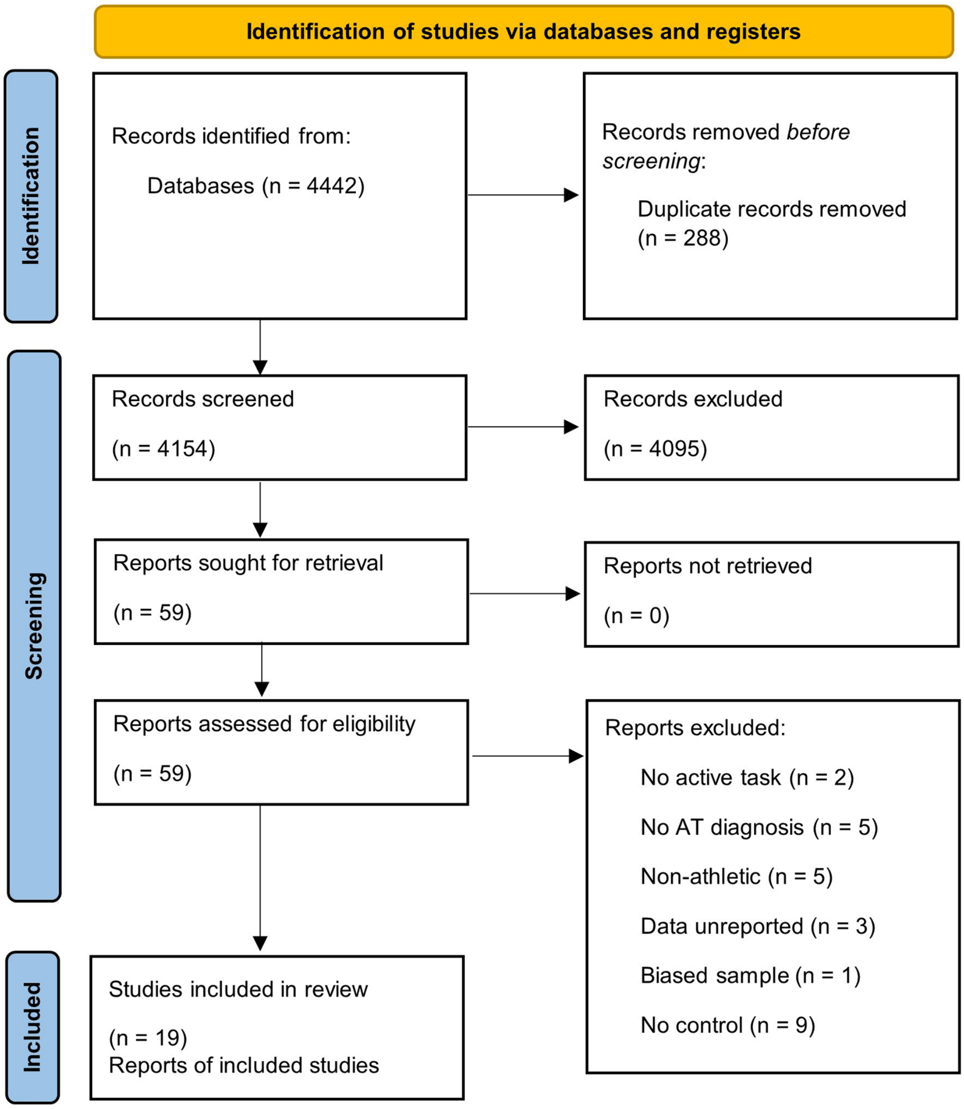

FIGURE 1
PRISMA flow diagram.

biomechanical variables during a "Non-gait multi-joint activity" (23, 43, 44), 8 of the studies investigated "Isolated joint strength" (32, 43, 45–50), and only 1 study explored "Reflex activity" (23). Nineteen of the included studies were in the English language, whereas one was written in German (48).

3.2. Participants

Details of the included studies can be found in **Table 2**, with information on participants, task characteristics, relevant outcomes, and calculated effect sizes with 95% confidence intervals. All studies included patients with Achilles tendinopathy (AT) and compared them to a healthy control group. A total of 769 participants were included across all 20 reports (19 experimental study populations). Seven studies included specifically male participants (34, 39–41, 43, 45, 48), a single study investigated females only (44), 9 experiments studied both males and females (32, 33, 36, 37, 42, 46, 47, 49, 50), and three studies did not report the sex of participants (23, 35, 38). The average age of all participants included was 38 years, with a large range from 18.5–50.5 years. Participants in the AT group were three years older on average across all studies (39.2 years vs. 36.2 years). All participants were considered athletic (ranging from recreational to elite), based upon inclusion criteria highlighted within the methodology. The majority of the included studies investigated runners exclusively ($n = 16$) (32, 33, 45–47, 34–40, 42), whereas three

TABLE 2 Study characteristics and results.

Authors, Study, Year & Study Type	Sample Group and number (n)/male-female/age [years (yr)]/mass (kg), height (cm), BMI (kg/m^2) Sporting activity [years (yr)]/training per week/ diagnosis (duration symptoms)/symptoms during testing	Task characteristics and category	Biomechanical variables	Significant findings (AT vs. CO) ($p < 0.05$) Relative direction of differences shown with arrow for AT group e.g., ↓ = decrease in variable compared to control	*P*-values	Calculated effect sizes (d) + (95% confidence intervals)
Baur et al. (2004) Case control	AT n = 8/NR/36 ± 9 yr/73 kg, 179 cm, NR Runners (NR)/5 ± 2 units/unilateral MPAT (>3 months)/symptomatic CO n = 14/NR/36 ± 9 yr/73 kg, 179 cm, NR Runners (NR)/5 ± 2 units	'Gait' Run/12 km/h/treadmill/Gymnastic barefoot shoe and standard running shoe	- Antero-posterior and vertical ground reaction forces - Plantar pressure distribution - EMG: Tibialis anterior, peroneus longus, gastrocnemius medialis, gastorcnemius longus and soleus	- Lateral deviation of centre of pressure whilst barefoot ↓	<0.001	−0.89 (−1.79, 0.03)
				- EMG amplitudes of extensor loop both barefoot and shod in weight acceptance ↓	<0.001	NA
Baur et al. (2011) Case control	AT n = 30/19m10f/41 ± 7 yr/72 kg, 175 cm, NR Runners (NR)/45 ± 21 km/unilateral MPAT (>3 months)/symptomatic CO n = 30/20m10f/37 ± 10 yr/67 kg, 174 cm, NR Runners (NR)/42 ± 14 km	'Gait' Run/12 km/h/treadmill Neutral running shoes	- EMG: tibialis anterior, peroneus longus and gastrocnemius medialis (Plantar pressure distribution used for calculation of touch down and take off time)	- Peroneus longus weight acceptance ↓	0.006	−0.54 (−1.05, −0.02)*
				- Gastrocnemius medialis weight acceptance ↓	0.001	−0.63 (−1.14, −0.10)*
				- Gastrocnemius medialis push-off ↓	0.04	−0.40 (−0.91, 0.12)
Becker et al. (2017) Case control	AT n = 13/9m4f/37.6 ± 15.9 yr/NR, NR, NR Runners (NR)/50.1 ± 15.1 miles/AT (NR)/ symptomatic CO n = 13/9m4f/32.6 ± 12.4 yr/NR, NR, NR Runners (NR)/52.32 ± 14.7 miles	'Gait' Run/Self-selected pace/5-m straight section Own running shoes	- 3D foot kinematics	- Rearfoot eversion heel-off ↑	<0.001	−1.63 (−2.51, −0.72)*
			- Ground reaction forces	- Period of pronation ↑	<0.001	1.72 (0.80, 2.62)*
Bramah et al. (2018) Case control (sub-group analysis)	AT n = 18/NR/38.5 ± 11.7 yr/63.1 ± 11.8 kg, 171.6 ± 8.7 cm, 21.3 ± 2.0 kg/m^2/ Runners (NR)/31.9 ± 17.6 miles/AT (>3 months)/ symptomatic (pain not >3 on NRS) CO n = 36/15m21f/33.2 ± 8.4 yr/60.8 ± 8.4 kg, 171.6 ± 7.3 cm, 20.6 ± 1.8 kg/m^2 Runners (NR)/60.5 ± 23.2 miles	'Gait' Run/3.2 m/s (11.5 km/h)/treadmill Own running shoes	- Kinematics at initial contact and mid-stance of trunk, pelvis, hip, knee and ankle.	- Contralateral pelvic drop midstance ↑	<0.01	1.37 (0.74, 1.99)*
				- Knee flexion initial contact ↓	<0.01	−0.87 (−1.46, −0.28)*
				- Ankle dorsiflexion initial contact ↑ (differences reported for pooled group of 4 separate injuries, but differences were consistent between injuries by sub-group analysis)	<0.01	0.72 (0.14, 1.30)*
Chang et al. (2015) Case control	AT n = 9/NR/46.8 ± 6.3 yr/74.1 kg, 170 cm, NR Runners, basketball, soccer, tennis (NR)/4,720 MET-min/AT (>2 wks)/asymptomatic (pain history)	'Non-Gait Multi-joint Activity' Hop/20 submaximal single leg hops/frequency 2.2 Hz/Own shoes	- Hopping: EMG - tibialis anterior, medial gastrocnemius, soleus and peroneal longus (Ground reaction forces for calculation of touch down and take off)	- "Individuals with Achilles tendinosis exhibit an earlier preactivation of the medial gastrocnemius on the involved side."	<0.001	3.42 (1.94, 4.85)*
	CO n = 10/NR/48.7 ± 4.4 yr/84.9 kg, 170 cm, NR NR/3,983 MET-min	'Reflex Activity' Electrical stimulation of tibial nerve and evoked muscle activity	- Reflex activity: Electromechanical delay, spinal and supraspinal responses	- "Meanwhile, the lowered contribution to plantar flexion from the triceps surae muscle was compensated for by other plantar flexors, such as the peroneal longus"	<0.001	−2.33 (−3.50, −1.12)*
				- "Both the H-reflex and the V-wave were higher on the side of Achilles tendinosis"	<0.001	H - 1.99 (0.85, 3.09)* V - 1.79 (0.69, 2.85)*

(*continued*)

TABLE 2 Continued

Authors, Study, Year & Study Type	Sample Group and number (n)/male-female/age [years (yr)]/mass (kg), height (cm), BMI (kg/m^2) Sporting activity [years (yr)]/training per week/ diagnosis (duration symptoms)/symptoms during testing	Task characteristics and category	Biomechanical variables	Significant findings (AT vs. CO) ($p < 0.05$) Relative direction of differences shown with arrow for AT group e.g., ↓ = decrease in variable compared to control	*P*-values	Calculated effect sizes (d) + (95% confidence intervals)
Child et al. (2010) Case control	AT $n = 14$/14m0f/40 ± 8 yr/80 ± 9 kg, 177 cm ± 6 cm, NR Runners (NR)/48 ± 18 km/MPAT (27 months)/symptomatic CO $n = 15$/15m0f/35 ± 9 yr/79 ± 11 kg, 178 cm ± 5 cm, NR Runners (NR)/42 ± 13 km	'Isolated Joint Strength' Maximal strength: isometric PF contraction with customized calf-raise apparatus with load cell/2-sec ramp followed by 3-sec MVC	- Isometric PF force (N)	- No significant differences between groups AT = 826.5 ± 246.8 N vs. CO = 755.6 ± 214.3 N	0.393	0.30 (−0.43, 1.03)
Creaby et al. (2017) Case control	AT $n = 14$/14m0f/43 ± 8 yr/82.3 kg, 179 cm, 25.73 kg/m^2 Runners (NR)/38.1 ± 13.2 km/MPAT (NR)/Symptomatic CO $n = 11$/11m0f/37 ± 9 yr/73.5 kg, 177 cm, 23.5 kg/m^2 Runners (NR)/35.9 ± 13.6 km	'Gait' Run/4 m/s (±10%)/25 m walkway Shoes: Nike Strap Runner IV	- 3D Kinematics: hip and ankle joints - Kinetics: joint moments of hip and ankle - Isometric plantar flexor peak torque (Nm)	- Hip external rotation at peak vGRF ↓	0.042	0.67 (−0.15, 1.48)
				- Hip peak external rotation joint moment ↑	0.0006	1.60 (0.67, 2.50)*
				- Hip external rotation impulse ↑	0.0001	1.50 (0.59, 2.39)*
				- Hip adduction impulse ↑	0.0005	1.67 (0.73, 2.58)*
Crouzier et al. (2020) Case Control	AT $n = 21$/18m3f/36.2 ± 8.3 yr/72.7 ± 8.7 kg, 175.9 cm ± 8.0 cm, 23.4 ± 1.5 kg/m^2 Mostly runners (NR)/3903 ± 2,105 MET·min·wk (2 h 36 min average of running per week)/ insertional ($n = 5$) and MPAT ($n = 16$) (3.6 months); 4 patients with bilateral symptoms/ Asymptomatic	'Isolated Joint Strength' Maximal Strength: MVIC of plantar flexors *via* isokinetic dynamometery, knee fully extended. 4 × 3 s.		- No differences in maxmal peak torque	0.403	−0.26 (−0.87, 0.35)
	CO $n = 21$/18m3f/35.1 ± 7.7 yr/71.8 ± 10.5 kg, 177.5 cm ± 7.8 cm, 22.7 ± 2.4 kg/m^2 Mostly runners (NR)/4601 ± 1,983 MET·min·wk (2 h 36 min average of running per week)	Submaximal strength: Isometric plantar flexor target torques of 20% and 40% of MVIC. 6 × 8 s trials at both intensities.	- Index of individual muscle (distribution of muscle force) % - EMG: Soleus, gastrconemius medialis, gastrocnmeius lateralis	- "We observed a different force distribution between the two populations with the GL muscle contributing significantly less to the overall submaximal isometric plantarflexion force in people with Achilles tendinopathy compared with the controls."	0.025	−0.54 (−1.15, 0.08)
Ferreira et al. (2020) Case control	AT $n = 25$/20m5f/36.84 ± 8.36 yr/73.90 ± 11.65 kg, 171 cm ± 6 cm, 24.97 ± 3.55 kg/m^2 Runners (6.7 ± 5.5 yrs)/36 ± 16.6 km/ MPAT & IAT mix (NR)/Asymptomatic CO $n = 26$/21m5f/35.07 ± 9.36 yr/73.55 ± 11.06 kg, 169 cm ± 7 cm, 25.4 ± 2.4 kg/m^2 Runners (6.2 ± 5.8 yrs)/40.15 ± 21.27 km	'Isolated Joint Strength' Isometric Maximal Strength (MVIC) of plantars flexors and hip external rotators using handheld dynamometer setup. 4 × 5 s MVIC, with 30 s rest.	- Isometric torque (Nm/Kg) for ankle plantarflexors and hip external rotators. - Hip internal rotation ROM and shank-forefoot alignment.	- "Runners with hip IR ROM under 13.99°, ankle PF torque above 0.76 Nm/kg, SFA above 5.53° and hip ER torque above 0.61 Nm/kg have higher chance of having AT." - "with hip IR ROM ≤29.33°, ankle PF torque >0.76 Nm/kg, and SFA >5.53°, hip ER torque was added into the model with a cut-off point of 0.61 Nm/kg. In this case, runners with hip ER torque below this value were identified with AT."	<0.001	PF: 0.21 (−0.34, 0.76) Hip: −0.28 (−0.83, 0.27)
Franettovich et al. (2014) Case control	AT $n = 14$/14m0f/43 ± 8 yr/82.3 kg, 179 cm, NR Runners (NR)/38.1 ± 13.2 km/MPAT (NR)/ Symptomatic CO $n = 19$/19m0f/37 ± 8 yr/77.4 kg, 179 cm, NR Runners (NR)/37.6 ± 16.4 km	'Gait' Run/4 m/s (±10%)/25 m walkway Shoes: Nike Strap Runner IV	- EMG: Gluteus medius and gluteus maximus (vertical ground reaction forces used to identify heel strike and toe-off)	- Gluteus medius onset delay	<0.001	2.11 (1.23, 2.96)*
				- Gluteus medius activity duration ↓	<0.001	2.31 (1.40, 3.19)*

(*continued*)

TABLE 2 Continued

Authors, Study, Year & Study Type	Sample Group and number (n)/male-female/age [years (yr)]/mass (kg), height (cm), BMI (kg/m^2) Sporting activity [years (yr)]/training per week/ diagnosis (duration symptoms)/symptoms during testing	Task characteristics and category	Biomechanical variables	Significant findings (AT vs. CO) ($p < 0.05$) Relative direction of differences shown with arrow for AT group e.g., ↓ = decrease in variable compared to control	*P*-values	Calculated effect sizes (d) + (95% confidence intervals)
				- Gluteus maximus onset delay	0.008	1.41 (0.62, 2.17)*
				- Gluteus maximus activity duration ↓	0.002	1.82 (0.98, 2.63)*
				- Gluteus maximus offset early	0.001	1.50 (0.71, 2.27)*
Habets et al. (2016) Case control	AT n = 12/12m0f/51.5 (43.0–53.0) yr/80.5 (75.0–92.9)kg, 189.0 (181.3–192.0)cm, 22.7 (21.8–26.8)kg/m^2 Runners, soccer, volleyball and tennis (34 yrs)/<3 h–>7 h/unilateral MPAT (17.5 months)/NR	'Isolated Joint Strength' Isometric maximal strength (MVIC) of hip abductors, hip external rotators and hip extensors, measured with HHD.	- Isometric strength (N/KgBw): Hip abductors, hip external rotators and hip extensors.	- Hip abductor strength ↓	0.012	NA
	CO n = 12/12m0f/49.5 (42.0–53.5) yr/80.5 (69.5–88.9)kg, 181.5 (179.3–185.8)cm, 23.7 (21.7–26.2)kg/m^2 Runners, soccer, volleyball and tennis (40 yrs)/<3 h–>7 h	'Non-Gait Multi-joint Activity' Functional hip performance - single leg squat (barefoot and underwear)	- Subjective rating of single-leg squat, based on "movement quality" criteria.	- Hip external rotator strength ↓	0.010	
				- Hip extension strength ↓	0.034	
				- No significant difference in functional hip performance		
Hein et al. (2013) Prospective	AT n = 10/8m2f/45 ± 5 yr/72 kg, 177 cm, 23 kg/m^2 Runners (NR)/33 ± 15 km/AT/ asymptomatic (healthy participants)	'Gait' Run/12 km/h/13 m runway Barefoot 'Isolated Joint Strength' Isometric strength: Hip abduction/adduction and knee flexion/extension	- Kinematics of hip, knee and ankle joints	- "As maximal joint excursions correlate with initial joint angles, it can be concluded that AT also show a more extended knee joint, a lower dorsiflexed ankle joint and a more everted rearfoot at touchdown compared with CO."	NR	Knee: −0.58 (−1.47, 0.33) DF: −1.21 (−2.16, −0.24)* EVR: −0.57 (−1.45, 0.34)
	CO n = 10/8m2f/40 ± 7 yr/72 kg, 177 cm, 23 kg/m^2 Runners (NR)/32 ± 20 km		- Maximal torque (Nm)	- "Runners who developed AT already showed decreased knee flexor strength compared with CO in an uninjured state even though 95% confidence intervals slightly overlap. No differences in maximal isometric strength were found for the hip joint surrounding muscles, or knee extensors between AT and CO."	NR	−0.91 (−1.83, 0.02)
Hirschmüller et al. (2005) Case control	AT n = 72/72m0f/39.4 ± 6.3 yr/74.0 ± 7.9 kg, 178.2 ± 5.7 cm, NR Runners (13.8 ± 8.7)/49.6 ± 5, 2 km/unilateral MPAT (NR)/NR	'Isolated Joint Strength' Maximum concentric and eccentric strength of ankle plantar flexion and dorsi flexion, isokinetic dynamometry.	- Maximum concentric and eccentric torque of ankle plantar flexion and dorsi flexion, measured with isokinetic dynamometry (velocity: 60°/s).	- Plantar flexor torque (Nm) both concentric and eccentric ↓	<0.05	NA
	CO n = 20/20m0f/28.7 ± 7, 9 yr/72.7 ± 9.6 kg, 180.6 cm ± 6.1 cm, NR Runners (8.2 ± 4.6)/37.0 ± 12.7		- EMG: Tibialis anterior, gastrocnemius medialis, gastrocnemius lateralis and soleus.	- EMG amplitudes gastrocnemius medialis, gastrocnemius lateralis and soleus during PF ↑	<0.01	
			- Neuromuscular efficiency quotient	- Neuromuscular efficiency ↓	<0.01	

(*continued*)

TABLE 2 Continued

Authors, Study, Year & Study Type	Sample Group and number (*n*)/male-female/age [years (yr)]/mass (kg), height (cm), BMI (kg/m^2) Sporting activity [years (yr)]/training per week/ diagnosis (duration symptoms)/symptoms during testing	Task characteristics and category	Biomechanical variables	Significant findings (AT vs. CO) ($p < 0.05$) Relative direction of differences shown with arrow for AT group e.g., ↓ = decrease in variable compared to control	*P*-values	Calculated effect sizes (d) + (95% confidence intervals)
Kulig et al. (2011) Case Control	AT $n = 8$/0m8f/18.5 ± 1.1yr/57.3 kg, 164 cm, NR Dancing (NR)/30 to 35 h/AT (>3 months)/asymptomatic (history of pain)	'Non-Gait Multi-joint Activity' Hop/saut de chat jump (maximal effort/height jump common in ballet) - minimum 3 trials	- Kinematics: Hip, knee and ankle joints.	- Hip adduction during braking phase ↑	0.046	1.04 (−0.03, 2.07)
	CO $n = 8$/0m8f/19.0 ± 1.3 yr/54.3 kg, 164 cm, NR Dancing (NR)/30 to 35 h	Without shoes		- Knee internal rotation during push-off phase ↑	0.024	1.25 (0.15, 2.31)*
Masood et al. (2014) Case Control	AT $n = 11$/7m4f/28 ± 4 yr/66 ± 6 kg, 174 cm ± 6 cm, NR Running, long jump, high jump, ice hockey (NR)/ 4.7 units/AT (9.8 months)/Asymptomatic	'Isolated Joint Strength' Maximal plantar flexor strength: isometric contraction *via* an in-house custom-built portable force transducer MVIC 8 sets of 5 unilateral,	- MVIC PF force, N 30% MVIC, N	- "Normalized myoelectric activity of soleus was higher ($P < 0.05$) in the symptomatic leg vs. the contralateral and control legs despite lower absolute force level maintained ($P < 0.005$).	<0.005	1.4* (NA)
	CO $n =$ 11/7m4f/28 ± 4yr/67 ± 6 kg, 173 cm ± 6 cm, NR Running, long jump, high jump, ice hockey (NR)/2.4 units	isometric, submaximal contractions at 30% of MVIC.	- EMG: Soleus, gastrocnemius medialis, gastrocnemius lateralis and hallucis longus	- No difference for maximal PF force between AT and CO		
O'Neill et al. (2019) Case control	AT $n = 39$/34m5f/47 ± 11.8yr/77 ± 12.1 kg, 177 cm ± 6.8 cm, 24 ± 2.7 kg/m^2/ Endurance runners (NR)/15–30 miles/unilateral MPAT (>3 months)/Asymptomatic	'Isolated Joint Strength' Maximal plantar flexor strength: measured by isokinetic dynamometry. Tested knee full extension and knee flexion 80° – concentric 90°/sec, concentric 225°/sec and eccentric 90°/sec.	- Peak plantar flexor torque (Nm & % body weight)	- "The results clearly show that there are large deficits in strength between subjects with and without AT. The magnitude of deficits is clinically and statistically significant in all test modes and both knee positions."	<0.001	Range: −0.83 (−1.30, −0.36*, −1.79 (−2.31, −1.25)*
	CO $n = 38$/35m3f/44 ± 9.9yr/70.4 ± 10.3 kg, 175 cm ± 8.1 cm, 23 ± 2.7 kg/m^2/ Endurance runners (NR)/15–30 miles	Plantar flexor endurance: 20 maximal effort concentric-eccentric plantar flexor contractions, positioned knee flexion 80°.	- Endurance capcity plantar flexors (Total Work Done)	- "The small percentage difference in force (healthy control torque/AT torque) observed between knee flexion and extension suggests that the gastrocnemius accounts for between 3.7%–11% of the identified deficits, whilst the soleus may be responsible for the remaining 23.2%–36.1% of the difference"		−1.25 (−1.73, −0.76)*
				- "The endurance data shows a clear clinically meaningful and statistically significant difference between subjects with and without AT"		
Ryan et al. (2009) Case control	AT $n = 27$/27m0f/40 ± 7yr/78 kg, 181 cm, NR Runners (>6 months)/>30 km/ AT (27 months)/symptomatic	'Gait' Run/pace self-selected/15 m runway	- Kinematics: Ankle (frontal and sagittal plane), tibia (transverse plane)	- Sub-talar joint eversion (mid-stance) ↑	0.04	0.67 (0.08, 1.25)*
	CO $n = 21$/21m0f/40 ± 9yr/71 kg, 177 cm, NR Runners (>6 months)/>30 km	Without shoes		- No difference in transverse tibial motion		
Van Ginckel et al. (2009) Prospective	AT $n = 10$/2m8f/38 ± 11.35yr/69.8 kg, 167,1 cm, 24.95 kg/m^2 Runners (Novice)/7.7 ± 1.03 h (walk + run)/AT/asymptomatic (healthy participants)	'Gait' Run/Self-selected speed/15 m runway Barefoot	- Plantar pressure force distribution	- Total posterior–anterior displacement of the Centre Of Force ↓	0.015	−0.95 (−1.65, −0.25)*
	CO $n = 53$/8m45f/40 ± 9yr/69.95 kg, 168.34 cm, 24.69 kg/m^2 Runners (Novice)/7.8 ± 1.24 h (walk + run)			- Laterally directed force distribution underneath the forefoot at "forefoot flat" ↑	0.016	−0.93 (−1.62, −0.23)*

(*continued*)

TABLE 2 Continued

Authors, Study, Year & Study Type	Sample Group and number (n)/male-female/age [years (yr)]/mass (kg), height (cm), BMI (kg/m^2) Sporting activity [years (yr)]/training per week/ diagnosis (duration symptoms)/symptoms during testing	Task characteristics and category	Biomechanical variables	Significant findings (AT vs. CO) ($p < 0.05$) Relative direction of differences shown with arrow for AT group e.g., ↓ = decrease in variable compared to control	*P*-values	Calculated effect sizes (d) + (95% confidence intervals)
Williams et al. (2008) Case control	AT $n = 8$/6m2f/36.0 ± 8.2yr/67.3 kg, 176 cm, NR Runners (19.1 ± 7.7 yrs)/41.3 ± 20.8 km/AT/ Asymptomatic (pain history) CO $n = 8$/5m3f/31.8 ± 9.3yr/65.6 kg, 170 cm, NR Runners (11 ± 9.1 yrs)/35.3 ± 23.1 km	'Gait' Run/3.35 m/s (± 5%)/20 m runway Saucony shoes/rear foot strikers	- Kinematics and joint moments in transverse plane of knee and tibia	- Tibial external rotation moment stance ↓	0.01	1.36 (0.24, 2.44)*
				- Peak knee internal rotation stance ↓	0.05	−0.97 (−2.00, 0.09)
Wyndow et al. (2013) Case Control	AT $n = 15$/15 m0f/42 ± 7yr/80 kg, 177 cm, NR Runners (NR)/>20 km; maximum of 43 ± 15 km/AT (>3 months)/symptomatic CO $n = 19$/19m0f/36 ± 8yr/77 kg, 179 cm, NR Runners (NR)/>20 km; maximum of 40 ± 16 km	'Gait' Run/4 m/s/25 m runway Shoes: Nike strap runners	- EMG: Soleus, gastrocnemius lateralis, gastrocnemius medialis (Ground reaction forces utilised to identify heel strike and toe-off)	- Earlier offset of soleus EMG activation relative to gastrocnemius lateralis	0.02	−0.90 (−1.60, −0.18)

*Statistical significance; NR, not reported; NA, not available; AT, achilles tendinopathy/participant with achilles tendinopathy; CO, control group; MPAT, mid-portion achilles tendinopathy; NRS, numerical rating scale; MET, metabolic equivalents; IAT, insertional achilles tendinopathy; Km, kilometres; Cm, centimetres; m, metre; m, male; f, female; BMI, body mass index; Kg, kilograms; Hrs, hours; Mins, minutes; Wk, week; Km/h, kilometres per hour; Secs, seconds; PF, plantar flexion; MVC, maximum voluntary contraction; MVIC, maximum voluntary isometric contraction; HHD, handheld dynamometry; EMG, electromyography; Hz, hertz; 3D, three dimensional; Nm, Newton metres; N, Newtons; ROM, range of motion; vGRF, vertical ground reaction force; IR, internal rotation; ER, external rotation; SFA, Shank-forefoot alignment; DF, dorsiflexion; EVR, eversion.

studies included running and other sports, such as basketball, soccer, tennis, volleyball, long jump, high jump and ice hockey (23, 43, 49), and a single study examined female dancers only (44). Methods of diagnosis for AT varied substantially, whereby ten studies identified "Achilles tendinopathy" (23, 32–34, 37, 38, 41, 42, 44, 49), five studies diagnosed unilateral "mid-portion Achilles tendinopathy" (35, 36, 43, 48, 50), three studies identified "mid-portion Achilles tendinopathy" without reference to side (39, 40, 45), and two studies included patients with both "insertional and mid-portion Achilles tendinopathy" (46, 47). Twelve of the 20 studies reported AT symptoms duration, and the range of duration was large (>2 weeks–27 months).

3.3. Outcomes

3.3.1. Gait activities

All 11 studies examining gait (32, 33, 42, 34–41) investigated running, at a variety of speeds and under different shod/barefoot conditions (see **Table 2**). None of the included studies researched other forms of human gait e.g., walking.

3.3.1.1. Kinematics

One prospective study investigated the kinematics of the hip, knee and ankle joints in twenty participants during running (32), and concluded that a more extended knee joint, a decreased angle of dorsiflexion at the ankle joint and a more everted rearfoot at touchdown preceded onset of AT. The remaining five studies investigated kinematics cross-sectionally. Four studies investigated ankle kinematics during running (37–39, 41). One study (37) reported changes in ankle kinematics with AT, including increased rearfoot eversion at heel-off and an increased period of pronation. Another study (41) also showed increased sub-talar joint eversion at mid-stance but no differences in ankle sagittal plane kinematics. One study (38) reported that AT patients exhibit increased ankle dorsiflexion at initial contact, but no differences in rearfoot frontal plane kinematics. Another study (39) showed no changes in sagittal nor frontal plane ankle kinematics for AT patients compared to healthy controls. One study (41) reported no difference in transverse tibial motion. A single study (38) reported reduced knee flexion at initial contact in the AT group, but no changes in knee kinematics in either the sagittal, transverse or frontal planes during midstance. A different study (42) showed a decrease in peak knee internal rotation angles within the AT group. One study (38) reported increased contra-lateral pelvic hip drop in the AT group compared to controls. A separate study (39) showed a difference in hip kinematics, namely a reduction in hip external rotation at peak ground reaction force (GRF), but not for four other variables in the sagittal and frontal planes.

3.3.1.2. Joint moments

Only two included studies investigated joint moments during running (39, 42). One study (39) reported no differences in ankle joint moments between groups, but demonstrated a decreased hip peak external rotation joint moment, hip external rotation impulse and hip adduction impulse in the AT group, with no differences in the sagittal plane. Another study (42) indicated a decreased tibial external rotation moment during stance phase, with no differences in transverse plane knee joint moment.

3.3.1.3. Ground reaction force

Two studies (35, 37) reported on GRF during running. Neither of these studies indicated any differences in vertical or propulsive and braking GRFs between the AT and a healthy control group.

3.3.1.4. Plantar pressure force distribution

A single prospective study (33) reported a significant decrease in posterior–anterior displacement of the centre of force and a laterally directed force distribution underneath the forefoot at "forefoot flat" during running, indicating AT onset in a prospective study design. Another study investigated plantar pressure force distribution cross-sectionally during running (35), showing a decreased lateral deviation of the centre of pressure in relation to the midline of the foot in AT whilst running barefoot.

3.3.1.5. Muscle activity (EMG)

A total of four studies investigated electromyographic changes (EMG) during running (34–36, 40). *Gastrocnemius:* One study (35) reported decreased amplitudes of the gastrocnemius lateralis during weight acceptance in the AT group, with no differences reported in timing. Another study (36) indicated reduced amplitudes of the gastrocnemius medialis during weight acceptance and push-off. A separate study (34) showed reduced offset EMG timing of the soleus relative to lateral gastrocnemius, although five other variables relating to muscle activity timing of the calf complex were not statistically significant. *Soleus:* One study (34) reported earlier offset of the soleus relative to gastrocnemius, whereas another study did not report any differences between groups (35). *Peroneus longus:* Two studies (35, 36) reported no differences in peroneus longus activity during pre-activation, although one of these studies (36) did show decreased activity during weight acceptance within the AT group. *Tibialis anterior:* Both studies (35, 36) reported no differences in tibialis anterior EMG activity between the AT group and controls. *Hip muscles:* One study (40) investigated EMG in muscles of the hip and reported a delayed gluteus medius onset, reduced gluteus medius activity duration, delayed onset of gluteus maximus, reduced gluteus maximus activity duration and earlier offset of gluteus maximus in the AT group.

3.3.2. Non-gait multi-joint activities

3.3.2.1. Sub-maximal hopping

One study (23) investigated hopping, finding that the AT group had a relatively lower contribution of the gastrocnemius and soleus muscles, compensated for by increased peroneus longus activity as measured by EMG.

3.3.2.2. Maximal jump

A single study (44) investigated the "saut de chat" ballet jump in dancers, and reported increased hip adduction during braking phase and increased knee internal rotation during push-off phase in the AT group, as measured by 3D kinematics of the hip, knee and ankle.

3.3.2.3. Functional hip performance

One study (43) subjectively assessed the function of the hip based upon pre-defined "movement quality" criteria during a single-leg squat, and reported no differences between the AT and control group in the subjective visual rating of postural stability and movement execution. The rating was based upon movement quality criteria in five domains and was subjectively rated by the investigators *via* video analysis in post-processing, whereby ratings for the domains were categorised as "poor", "fair", or "good" and then indexed into a total score.

3.3.3. Isolated joint strength

Eight of the studies investigated "Isolated joint strength" (32, 43, 45–50), and reported on a wide range of biomechanical strength variables. Measurement techniques varied, including isokinetic dynamometry (46, 48, 50), handheld dynamometry (43, 47) and other custom made devices (32, 45, 49). Subject positioning also differed between studies to a large degree, depending on apparatus used and the joint of interest. Six studies reported on strength of the ankle joint (45–50), one study investigated the knee joint (32) and three studies reported on the hip joint (32, 43, 47).

3.3.3.1. Maximal strength

Only one study of twenty subjects (32) investigated maximal isometric strength prospectively, and identified decreased knee flexor strength in runners who went on to develop AT. No differences in maximal isometric strength were found for the hip joint surrounding muscles, or knee extensors between AT and control subjects. Regarding cross-sectional study designs, a total of six studies investigated maximal strength of the ankle joint (45–50). Two studies (48, 50) found associations between reduced maximal plantar flexor (PF) strength in the AT group, during both concentric and eccentric muscle contractions on an isokinetic dynamometer. In one of these studies (50), the effort was produced with the knee both fully extended and bent at 80°. One other study (47) reported that increased isometric PF strength was associated with AT, but only when associated with other biomechanical factors. Three different studies (45, 46, 49) discovered no differences in isometric PF strength between the AT group and heathy controls. One study (43) investigated isometric maximal strength of the hip, and reported reduced hip abductor strength, reduced hip external rotator strength and decreased hip extension strength in the AT group. A separate study (47) reported that both increased and decreased isometric hip external rotation strength when combined with other biomechanical factors, were associated with AT.

3.3.3.2. Strength endurance

One study (50) investigated plantar flexor endurance (20 repetition protocol) *via* isokinetic dynamometry, and reported significant and clinically meaningful deficits in the AT group compared to healthy controls.

3.3.3.3. Muscle activity (EMG)

A total of three studies (46, 48, 49) investigated muscle activity during isolated joint strength activities. All studies measured strength of the ankle joint in plantar flexion, whilst one study also measured dorsi flexion (48). One study (46) measured soleus, gastrocnemius medialis and gastrocnemius lateralis EMG activation, and reported a lower contribution of gastrocnemius lateralis activity to overall triceps surae output in the AT group, during sub-maximal intensities [20% and 40% of maximum voluntary isometric contraction (MVIC)]. Two studies showed increased EMG activity of the soleus (49), and soleus, gastrocnemius medialis and gastrocnemius lateralis muscles (48) within AT patients, despite lower levels of overall plantar flexor force output in both studies.

3.3.4. Reflex activity

Only a single study (23) investigated reflex activity. The authors reported an up-regulated spinal reflex at rest (H-reflex) and accentuated supraspinal reflex responses (V-Wave) during MVIC, on the involved side of AT patients when compared to healthy controls.

3.4. Risk of bias

The risk of bias assessment according to the Critical Skills Appraisal Checklist can be seen in **Table 3**. Overall, it could be concluded that the included studies scored poorly in "Appropriate Recruitment of Controls", "Control of Confounding Factors" and "Generalizability of the Results". There was a wide variation of methodological approaches applied, especially regarding recruitment of controls, symptoms of injury, footwear, positioning of the participants, the measurement techniques utilised and the statistical designs of the studies. Therefore, it could be concluded that risk of bias in the included studies was predominantly moderate/high.

4. Discussion

This study aimed to synthesise the evidence regarding biomechanical alterations and changes in lower limb function in patients with AT. The included studies investigated exclusively athletic populations, from recreational to elite level, when compared to a healthy athletically matched control group. Throughout the discussion, effect sizes and statistical significance are reported from the relevant studies, to allow for clear comparison and interpretation (reported as: Effect Size (ES) d [95% Confidence Intervals (CIs)]. Statistical significance is indicated by an asterisk*). Overall, it must be emphasised that most of the reported biomechanical variables produced conflicting results within this review, especially regarding the data during running and jumping activities. Whilst a number of biomechanical theories have been postulated elsewhere in the literature, for example "medial collapse theory" and "contralateral pelvic hip drop" (20, 38), the authors of the current study do not believe that there is sufficient evidence to support or refute any such theories based upon the research summarised within this review. We would explicitly recommend against drawing concrete conclusions and applying them on absolute terms in clinical practice, until more evidence has been gathered and the picture is clearer. Nonetheless, readers will find an attempt to interpret and discuss the data collected in this review, in the context of popular theories within the realm of sports biomechanics. This should in no way be considered as an endorsement of these theories or approaches.

4.1. Potential biomechanical alterations during gait

There is some evidence to suggest that ankle biomechanics may be altered in athletic AT patients, although the results were conflicting, and several variables were not associated with AT. Increased ankle eversion during running was correlated with injury both prospectively [$d = -0.57$ (−1.45, 0.34)] and cross-sectionally (37, 41) ($d = -1.63$ [−2.51, −0.72]*; $d = 0.67$ [0.08, 1.25]*). An increased period of pronation during running was also found to identify patients with AT (37), with strong ES [$d = 1.72$ (0.80, 2.62)*]. This data potentially corroborates previous suggestions that over-pronation of the foot may produce a "whiplash effect" (51), placing excessive strain on the Achilles tendon and leading to injury. This theory is further supported by evidence of increased medial deviation of the foot whilst running, as measured by plantar pressure distribution (35) [$d = -0.89$ (−1.79, 0.03)]. However, in two of these studies, 95% CIs of the ES overlapped zero (32, 35), indicating less statistically robust results. The crossing of the zero was nevertheless particularly small in one study [0.03 (32)], so this

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TABLE 3 Risk of bias assessment for included studies.

Study and year	Focused Issue?	Appropriate Methodology?	Apprpriate Recruitment of Patients?	Appropriate Recruitment of Controls?	Exposure Accurately Measured?	Control of Confounding Factors?	Precision of Results?	Statistics Clearly Reported?	Generalizability of Results?	Results Fit With Other Evidence?
Baur et al. (2004)	Yes	No	No	No	Yes	No	Yes	Yes	No	Yes
Baur et al. (2011)	Yes	Yes	Yes	No	Yes	No	No	Yes	No	Yes
Becker et al. (2017)	No	Yes	No	No	No	No	Yes	No	No	Yes
Bramah et al. (2018)	No	Yes	Yes	Yes	Yes	No	No	No	Yes	Yes
Chang et al. (2015)	Yes	Yes	Yes	Yes	Yes	No	No	Yes	No	Yes
Child et al. (2010)	Yes	Yes	No	No	Yes	No	Yes	Yes	Yes	Yes
Creaby et al. (2017)	Yes	Yes	Yes	No	Yes	No	Yes	Yes	No	No
Crouzier et al. (2020)	Yes	Yes	Yes	No	Yes	No	Yes	No	No	Yes
Ferreira et al. (2020)	No	No	No	No	Yes	No	No	Yes	No	No
Franettovich et al. (2014)	Yes	Yes	Yes	No	Yes	No	Yes	Yes	No	Yes
Habets et al. (2016)	Yes	No	Yes	No	No	No	No	Yes	No	Yes
Hein et al. (2013)	Yes	Yes	No	No	Yes	No	No	No	No	Yes
Hirschmüller et al. (2005)	Yes	Yes	Yes	No	Yes	No	Yes	Yes	No	Yes
Kulig et al. (2011)	Yes	No	Yes	Yes	Yes	No	No	No	No	Yes
Masood et al. (2014)	Yes	Yes	Yes	No	No	Yes	No	No	No	No
O'Neill et al. (2019)	Yes	Yes	Yes	No	Yes	Yes	No	Yes	No	Yes
Ryan et al. (2009)	Yes	No	Yes	Yes	Yes	No	Yes	No	No	Yes
Van Ginckel et al. (2009)	Yes	Yes	Yes	Yes	Yes	No	Yes	No	No	No
Williams et al. (2008)	Yes	Yes	Yes	No	Yes	No	No	No	Yes	Yes
Wyndow et al. (2013)	Yes	No	Yes	No	Yes	No	Yes	No	No	Yes

should be taken into account. It must also be considered that many studies indicated a large variation in effect sizes, perhaps signaling that these factors might be more relevant for some individuals than others, within the various groups studied. Furthermore, two other studies (38, 39) were not able to detect a difference between groups in transverse or frontal plane kinematics at the ankle. Additionally, a prospective study (33) reported an increase in the laterally directed force distribution at forefoot flat phase prior to the onset of AT in novice runners [$d = -0.93$ (−1.62, −0.23)*], contradicting the proposed over-pronation hypothesis. Besides, a closer look at absolute values of two studies (32, 41) reporting statistically significant differences in rearfoot eversion, reveals mean differences between the AT and control group of 2 degrees ankle eversion. Whether such changes are clinically detectable and/or meaningful, is a question requiring more attention and research. Perhaps these disparities are more representative of natural movement variability, as opposed to true biomechanical differences on reductionist terms (52, 53). Additionally, three of the included studies investigating ankle biomechanics had participants run shod (37–39), whereas the other three studies instructed subjects to run barefoot (32, 33, 41), and such a methodological discrepancy is likely to have influenced outcomes, especially in kinematics of the ankle joint. Finally, studies that only report on plantar pressure distributions during running (33) offer limited value, as the overall kinematic picture of foot loading is absent, and future research should aim to integrate both kinetic and kinematic measurements simultaneously.

Two studies reported no differences in sagittal plane ankle kinematics when comparing AT patients to healthy controls (39, 41). However, changes in sagittal plane ankle kinematics were reported in two other studies during running (32, 38), with one study showing increased dorsiflexion in the AT group (38) [$d = 0.72$ (0.14, 1.30)*], and a single prospective study associating decreased ankle dorsiflexion with onset of AT (32) [$d = -1.21$ (−2.16, −0.24)*], whereby ES are moderate to large in either direction. It could be speculated that a more compliant strategy at the ankle as seen in one study (38), meaning increased dorsiflexion range of movement (ROM), may result in higher loads on the Achilles tendon during running and potentially lead to injury (54, 55). However, it seems very difficult to support this hypothesis based upon current evidence, especially in light of findings suggesting that static dorsiflexion range of motion (ROM) might be reduced in AT (37), alongside prospective evidence associating decreased ankle dorsiflexion with onset of AT (32). Perhaps it could be interpreted that both increased and decreased dorsiflexion ROM might be associated with AT, depending upon individual factors. However, it could just be attributed to natural variability, and much more research is required before any conclusions can be made.

Three studies reported on electromyographic outcomes of the ankle muscles during running. Two of these studies showed decreased EMG amplitudes of the plantar flexor muscles during weight acceptance (35, 36) [$d = -0.63$ (−1.14, −0.10)*], whilst a single study indicated reduced activity of the gastrocnemius medialis during push-off phase (36), though with a small effect size and non-statistically significant 95%CIs [$d = -0.40$ (−0.91, 0.12)], although the zero was only crossed by a minimal degree (0.12). This may represent a diminished capacity in AT patients of the triceps surae and Achilles tendon unit to attenuate loads eccentrically, and to store and release energy efficiently during propulsion, as seen in healthy running (56). It could also be indicative of inhibitory processes due to the pain often associated with AT, which has been demonstrated to alter motor behavior (27, 57, 58). Additionally, another study (34) indicated altered temporal activation of the triceps surae muscles in patients with AT [$d = -0.90$ (−1.60, −0.18)*], providing further evidence of potentially compensatory adaptations to persistent injury (25, 57, 58). However, the data from only three studies is not sufficient to draw concrete conclusions, and further research is required.

The data reporting on variables of the hip during gait provides a conflicting picture, and it is very difficult to infer a coherent pattern. There is evidence from one study (38) to suggest that increased contra-lateral pelvic hip drop during running is associated with AT, with a large effect size and robust confidence intervals [$d = 1.37$ (0.74, 1.99)*]. This contrasts with other prospective research (32) highlighting no differences in kinematics of the hip prior to onset of AT. Another study (39) showed decreased hip external rotation ROM at peak vertical ground reaction force in AT, but 95%CIs crossed the value of zero effect [$d = 0.67$ (−0.15, 1.48)] which weakens the findings, although it was by a small degree (−0.15). The same study (39) additionally reported alterations in mechanics of the hip, reporting increased external rotation impulse and joint moments, and increased hip adduction impulse (large ES, see **Table 2**). In addition, research investigating EMG during running (40) indicates reduced duration and delayed onset of muscle activity in the gluteus maximus and gluteus medius (large ES, especially for gluteus medius, see **Table 2**) within AT patients compared to controls. Whether these proximal changes occur as a consequence of alterations in local ankle biomechanics relating to AT pathology, or are an isolated feature, is difficult to deduce based upon current evidence. Besides, the data is conflicting and only based upon four studies. It seems plausible that the potential adaptations associated with AT overuse injury throughout the kinetic chain are interrelated (20, 56), though the exact mechanism remains unknown. Alterations in hip biomechanics have been reported in other common running injuries (59), and there is emerging evidence to suggest that interventions targeting gait-retraining e.g., to alter proximal hip kinematics, paired with strengthening interventions, may have a beneficial effect on pain and function (14, 59). Whether such interventions are of true clinical benefit to AT patients requires further investigation. Moreover, most of the included studies were cross-sectional by design. Therefore, whether the above-mentioned biomechanical changes during running are true risk factors that occur prior to AT onset or are adaptations to the condition post-onset, is a question that remains elusive to answer.

4.2. Potential biomechanical alterations during non-gait functional activities

Only three studies investigated the biomechanics of AT patients in non-gait functional activities (23, 43, 44), which makes it challenging to draw overall conclusions. One study (23) detected strong effects of a lower contribution of the triceps surae muscles in AT during 20 sub-maximal hops, compensated for by increased peroneus longus activity [$d = -2.33$ (−3.50, −1.12)*]. These neuromuscular alterations agree with the evidence discussed for running studies (35, 36, 60), and perhaps represent a broad trend, whereby athletic patients with AT present with altered activation of the triceps surae muscles during stretch shortening activities such as running and hopping. Another study (44) found changes in hip and ankle kinematics in female dancers during a "saut de chat" single unilateral maximum jump (jump common in ballet). Effect size for increased hip adduction and increased knee internal rotation were strong but 95%CIs indicate a high level of variability amongst participants ($d = 1.04$ [−0.03, 2.07]; $d = 1.25$ [0.15, 2.31]*). These findings seem to confer with other results in this review (32, 35, 37–40) potentially indicating an overall biomechanical picture of "medial collapse" during dynamic loading of the lower-limb, featuring contralateral pelvic hip drop and increased hip adduction, knee valgus and increased internal rotation, ankle over-pronation and reduced capacity of the hip stabilisers, which may predispose people and/or be associated with the development of AT or other running injuries (61). This neuromechanical pattern is thought to be a particular risk factor in females, and this population might benefit most from interventions targeting these specific motor behaviours (14, 61). However, it must be stressed that the data is conflicting and at times contradictory. In fact, prospective evidence from one study reported lateral foot deviation as a risk factor for AT development (33), which certainly challenges the commonly purported "medial collapse" hypothesis. The evidence presented in this review is not strong enough to be conclusive and should direct future high quality research studies to replicate or reject the findings. Until then, our assumptions are merely based upon speculation. It should also be emphasised that several studies investigated these biomechanical variables and reported no differences between groups. A single study assessed a one-leg squat in AT compared to controls (43), and found no differences in functional hip performance according to standardised subjective criteria. It could be hypothesized that a single-leg squat does not demand stretch shortening activity of the Achilles tendon and kinetic chain of the lower limb, as opposed to running and hopping activities which are known to load the Achilles to a large degree and potentially lead to pathology (2, 8, 9). Although speculative, this might explain why no kinematic differences were found between the groups and perhaps represents a specific kinematic adaptation of AT patients when performing SSC movements, but not during closed chain squatting.

4.3. Reduced knee flexor strength prospectively

A single study (32) showed that reduced isometric knee flexor strength predicted onset of AT prospectively, with strong effects but wide 95%CIs [$d = -0.91$ (−1.83, 0.02)]. The zero is crossed in this instance, but only by a very small margin (0.02) and therefore, is potentially irrelevant. The importance of the hamstring muscles in load attenuation and propulsive sprint efforts is well documented in literature (62) and could form a potential treatment target for AT patients. However, data from a single study is not currently sufficient to make explicit recommendations.

4.4. Reduced maximal dynamic plantar flexor strength but not isometrically

Two studies investigated plantar flexor strength dynamically *via* isokinetic dynamometry (48, 50), with both studies showing reduced maximum concentric [$d = -1.25$ (−1.74, −0.76)*] (50), and eccentric torque [$d = -1.38$ (−1.88, −0.88)*]. Effect sizes could not be calculated for one of these studies (48) but strength deficits in the AT group were reported as between 10% and 20%, which is less than the 30%–40% deficits seen in absolute values in the comparative study (50). This difference might be explained by the older population included within the O'Neill et al. study (50), which could be correlated with longer duration of symptoms and therefore, exacerbated mechanical adaptations to prolonged pathology. When maximal strength values were normalised to body weight (kg) in one study (50) discrepancies became even more prominent for AT vs. control group, with deficits of 40%–45% reported for the extended and flexed knee positions, in both concentric and eccentric modes and showing strong effect sizes [Max. value: $d = -1.79$ (−2.31, −1.25)*]. This approach is an interesting avenue for future research and could be easily applied in a clinical setting when working with AT patients. The authors of this study (50) additionally postulate that the gastrocnemius muscle accounts for 3.7%–11% of deficits, whereas the soleus might account for between 23.2%–36.7% of plantar flexor strength deficits. This could specifically imply training of the soleus muscle as a rehabilitation strategy for AT patients, although whether a training intervention is able to specifically target the soleus is still debatable (63, 64). The O'Neill study (50) also identified significant and clinically meaningful deficits in muscular endurance of the plantar flexors of AT runners, which may be particularly relevant for populations involved in endurance sports requiring repeated loading of the tissues over long periods where fatigue is a factor. Interestingly, three of the four studies investigating isometric plantar flexor strength did not identify any differences between AT and healthy control groups (45, 46, 49). This suggests that isometric testing might not be sufficient to identify strength deficits within an active, athletic population, perhaps alluding to a specific adaptation of the musculotendon unit in AT pathology that does not affect maximal isometric force output. Therefore, despite existing evidence that isometric contractions may have a positive effect on pain and function in tendinopathies (58, 65), isotonic exercises should be considered for strength testing and rehabilitation as soon as symptoms allow. An additional study (47) reported that increased isometric plantar flexor strength was associated with AT, but only in combination with a number of other biomechanical factors when integrated within an interactive statistical model. A closer investigation of absolute values reveals only a 0.05 [Nm normalised to body weight (kg)] difference between groups and small ES with non-statistically significant 95%CIs [$d = 0.21$ (−0.34, 0.76)], bringing into question the clinical relevance of the results.

4.5. Alterations in triceps surae activity during plantar flexion

Three studies reported on muscle activity of the triceps surae muscles during isolated ankle plantar flexor strength in AT compared to control (46, 48, 49). All three studies showed differences between groups, with two studies (48, 49) highlighting an increase in triceps surae muscle activity, despite a decrease in overall force output within the AT group ($d = 1.4^*$) (49). This might indicate a reduced efficiency of the plantar flexors to generate force, in relation to AT pathology (48) or could be a consequence of pain inhibition and central factors (25, 57), whereby pain has been shown to reduce the force output and efficiency of the plantar flexors in healthy populations (66). However, these two studies had different methodological approaches, for example one study reported on maximal contractions (48) whereas the other study investigated sub-maximal contractions (49), therefore direct comparison between studies should be conducted with caution. A different study (46) found alterations in the force sharing profile of the triceps surae muscles in patients with AT vs. controls, reporting a reduced contribution of the lateral gastrocnemius muscle to sub-maximal isometric plantar flexion [$d = -0.54$ (−1.15, 0.08)]. Although, ES is moderate and 95%CIs do cross zero, which should promote caution, even though it is by a small amount (0.08). However, this finding supports results from other studies within this review, which detected changes in triceps surae activation during running (35, 60, 67) and hopping (23) activities in AT. It could be suggested that alterations in the electromyographic profile of the triceps surae muscle unit are apparent across a range of movement tasks, but that the exact nature of these changes and the causal mechanism requires further deliberation. Besides, the methodological quality of these studies is questionable, and future high-quality trials are necessitated.

4.6. Potential changes in hip strength for AT

One study (43) reported that isometric maximal strength of the hip abductors, extensors and external rotators was reduced in AT, with deficits ranging from 28.3%–34.2%. These muscles are the key stabilisers of the proximal limb segment, and a weakness could result in a redistribution of force absorption throughout the kinetic chain, perhaps leading to injury. Conflictingly, a different study (47) found that both increased and decreased maximal hip external rotation strength were associated with AT when associated with a range of other biomechanical variables within an interactive statistical model. Again, upon closer inspection of absolute values the deficits were not clinically meaningful with only small and non-significant effects [$d = -0.28$ (−0.83, 0.27)]. Additionally, a prospective study (32) was also unable to identify AT patients when considering hip abduction and adduction strength. Therefore, whether strength changes at the hip for athletic AT patients are relevant is difficult to conclude based upon evidence within this review. Although, it still would seem sensible to consider these factors within the clinical reasoning process, based upon other biomechanical alterations reported within AT. Overall, it could be concluded that rehabilitative strategies focusing on the restoration of plantar flexor strength, potentially hip strength in movements of extension, external rotation, and abduction, and possibly knee flexor strength should be incorporated within clinical practice when treating athletic AT patients. The exact mechanism by which such interventions benefit pain or function remains unclear.

4.7. Reflex activity upregulated in AT

Both spinal and supraspinal reflexes were reported to be upregulated within AT patients affected side compared to controls, but this was only found in a single study (23). This may indicate a protective response of the injured tendon, perhaps mediated by central factors (25–27). Whether the normalization of reflex responses should be targeted with interventions for AT patients, and their relation to pain and function, is a novel area of research requiring further investigation.

4.8. Limitations

The risk of bias assessment indicated a large variation between study designs and methodological approaches, meaning the results of this review should be interpreted cautiously and drawing any conclusions based upon the current data is extremely difficult. A key problem identified in many studies was that control groups were not matched to the patient group by age, whereby the control group was substantially younger in a number of cases (32, 34, 46–48, 50, 36, 37, 39–43, 45). This may have affected the amount of time spent training within a participant's individual sport, and/or the duration of AT pathology, which might directly impact the findings compared between groups. Besides, age is purported to be a risk factor in general for the development of tendinopathy (68). Data for training duration was only reported in five out of the twenty studies (41–43, 47, 48), making such comparisons difficult to conduct. Fourteen of the twenty studies were determined to have a high risk of bias regarding male vs. female sex inclusion, whereby seven of the studies included substantially more males than females within the study design (32, 36, 37, 42, 46, 47, 50) and a further seven studies only included male participants (34, 39, 41, 43, 45, 48, 69). This limits the generalisability of the results to the female population. The bias assessment also identified discrepancies in the symptomatic behaviour of patients within the AT group across studies. Nine of the included studies investigated patients presenting with current symptoms of pain (34–41, 45), whereas the other eleven studies only included AT patients who were currently without symptoms and in a period of remission. The effects of pain on motor behaviour are well documented (23, 25, 46), and should be considered when interpreting findings of the included studies. Finally, the included studies used a wide range of protocols to investigate parameters of gait, joint strength, and other movement behaviours. For example, some participants ran shod (34–40, 42) whilst others were barefoot (32, 33, 41). Moreover, various studies allowed running at a self-selected speed (32, 35, 36, 38–40, 42, 60) while others standardised a specific speed for all participants (33, 37, 41). Such variation in methodologies makes comparisons and discernment of concrete conclusions challenging. As a final point on the design of the included studies, the statistical reporting was not clear in many experiments (see **Table 3**) and several biomechanical variables were often tested for statistical significance on a single population. This might raise the chance of finding statistical significance by chance alone, and future studies should be designed with appropriate statistical models and be adequately powered. There are also some limitations to be acknowledged related to the methods of this review. Strict inclusion criteria were applied e.g., athletic population, AT diagnosis, healthy control group, meaning a large body of literature regarding AT and biomechanics could not be included for synthesis, and this research is cited here for transparency (63, 70–84). The main reasons for excluding these studies were the study of a non-athletic population or because relevant information could not be obtained from the authors. Whilst a limitation to some extent, this is also inherent to the strength of the study design. Studies were excluded so that a specific athletic population could be considered, in experiments which investigated AT patients compared to healthy control groups as opposed to the contralateral limb. Given research indicating sensory and motor deficits on the contralateral limb and altered pain processing within AT patients (25, 27) this approach seems justified, and potentially more effective in identifying biomechanical alterations or impairments within the AT population. Additionally, for two studies (43, 48) effect sizes couldn't be calculated as the data was unsuitable or unavailable.

4.9. Conclusions

According to evidence synthesised in this review, there appear to be notable biomechanical alterations during a range of movement tasks in athletic populations with AT compared to their healthy control group counterparts. Equally said, there were several biomechanical variables investigated that were not associated with AT, and in general the study quality of the included trials was poor. This is in agreement with other reviews of research in this area that investigated mixed athletic and general populations (2, 10, 20, 21). Having addressed several of the postulated theories regarding habitual motor patterns and their relationship with AT in this review, the authors would find it very difficult to either accept or refute their relevance based upon the current evidence, especially for those related to running gait kinematics. In summary, the proposed alterations include changes in kinematics and muscle activity of the hip and ankle joint during running, alterations in lower limb function during jumping/hopping, strength deficits of the plantar flexors, the knee flexors and possibly the hip joint, and weak evidence for up-regulated reflex activity. It seems logical to conclude that these alterations might form potential treatment targets for clinical interventions, for example strengthening programs for the kinetic chain of the entire lower limb with particular emphasis on the plantar flexors, knee flexors and hip, gait re-training, plyometrics to restore the stretch shortening capacity of the musculotendon unit, and possibly sensory motor training. However, much more research is required in longitudinal study designs before any concrete conclusions can be drawn from the data within this review. Additionally, the effectiveness and exact mechanisms of improvement with such interventions necessitates further research, and these treatments should be applied on an individual basis with consideration of the specific needs of each patient. It should also be emphasised that the biomechanical profile of Achilles tendinopathy patients is likely to be one of many contributing factors to the overall clinical picture, whereby other factors such as training load management, genetics, previous musculoskeletal injuries, cardiometabolic profile, BMI, psychosocial factors, and other co-morbidities, should also be considered. Although, one might expect factors such as training load, previous musculoskeletal injuries, and biomechanics to play a larger role in athletic populations. Future high quality prospective studies are required to explore the causal mechanisms of AT onset and its relation to biomechanics in athletic groups. Until such studies are conducted, it is very difficult to ascertain whether biomechanical variables are the cause or consequence of musculoskeletal injuries such as AT. The altered biomechanical variables reported in this review, could serve as a good starting point for the focus of such research investigations. If future high-quality trials can confirm these alterations, then clinicians might utilise these as clinical markers in the prevention and rehabilitation of Achilles tendinopathy. However, for the time being, caution is very much warranted and there are no solid conclusions that can be drawn based upon the evidence within this review, due to the reported low-quality of the research and paucity of investigations.

Data availability statement

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author.

Author contributions

AQ – Conceptualization; Data curation; Formal analysis; Investigation; Methodology; Project administration; Resources; Roles/Writing—original draft; Writing—review & editing. JM – Conceptualization; Data curation; Formal analysis; Investigation; Methodology; Resources; Writing—review & editing. HM – Conceptualization; Methodology; Data curation; Writing—review & editing. JH – Conceptualization; Data curation; Writing—review & editing. MHK – Conceptualization; Data curation; Writing—review & editing. Michael Cassel—Conceptualization; Writing—review & editing. TE – Conceptualization; Data curation; Investigation; Methodology; Project administration; Resources; Writing—review & editing; Project Supervision. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Publisher's note

Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2022.1012471/full#supplementary-material.

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Effect of acute ankle experimental pain on lower limb motor control assessed by the modified star excursion balance test

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Introduction: Following most musculoskeletal injuries, motor control is often altered. Acute pain has been identified as a potential contributing factor. However, there is little evidence of this interaction for acute pain following ankle sprains. As pain is generally present following this type of injury, it would be important to study the impact of acute pain on ankle motor control. To do so, a valid and reliable motor control test frequently used in clinical settings should be used. Therefore, the objective of this study was therefore to assess the effect of acute ankle pain on the modified Star Excursion Balance Test reach distance.

Methods: Using a cross-sectional design, 48 healthy participants completed the modified Star Excursion Balance Test twice (mSEBT1 and mSEBT2). Following the first assessment, they were randomly assigned to one of three experimental groups: Control (no stimulation), Painless (non-nociceptive stimulation) and Painful (nociceptive stimulation). Electrodes were placed on the right lateral malleolus to deliver an electrical stimulation during the second assessment for the Painful and Painless groups. A generalized estimating equations model was used to compare the reach distance between the groups/conditions and assessments.

Results: *Post-hoc* test results: anterior ($7.06 \pm 1.54\%$; $p < 0.0001$) and posteromedial ($6.53 \pm 1.66\%$; $p < 0.001$) directions showed a significant reach distance reduction when compared to baseline values only for the Painful group. Regarding the anterior direction, this reduction was larger than the minimal detectable change (5.87%).

Conclusion: The presence of acute pain during the modified Star Excursion Balance Test can affect performance and thus might interfere with the participant's lower limb motor control. As none of the participants had actual musculoskeletal injury, this suggests that pain and not only musculoskeletal impairments could contribute to the acute alteration in motor control.

KEYWORDS
motor control, pain, ankle, star excursion balance test, SEBT, ankle sprain

1. Introduction

Ankle sprains are frequent musculoskeletal (MSK) injuries (1–3). After an initial ankle sprain, approximately 33% of patients suffer from chronic ankle instability (4), reporting residual symptoms such as recurrent sprain, episodes of ankle joint "giving way," pain, swelling, and decreased function (5). Chronic ankle instability can be perceived up to 3 years following the injury (4). Moreover, up to 78% of the individuals with ankle injuries are at risk of developing ankle osteoarthritis (6, 7). Therefore, adequate follow-up of people who sustained an ankle sprain is crucial to prevent chronicization and further damage at the ankle.

A wide variety of tests has been developed to assess individuals with ankle injuries. These tests can either assess somatosensation or motor control (8). Somatosensation tests imply the use and interpretation of sensitive information from sensory receptors such as muscle spindles, Golgi tendon organs, joint receptors and cutaneous receptors from skin over the joints (9). Somatosensation tests are useful following ankle sprain as this type of injury can further alter somatosensation (10). Motor control tests give information about the ability to regulate or direct the mechanisms essentials to movement (11), thus assessing performance during functional movement execution. A recent systematic review reported that the Star Excursion Balance Test (SEBT) is the most valid, reliable and responsive test to assess the lower limb motor control of individuals with a sprained ankle (8). Initially described with a participant standing on one leg and reaching as far as they can on a star-shaped form (12), this motor control test also has two short versions using a Y-shaped form showing similar psychometric properties, the modified Star Excursion Balance Test (mSEBT) (13) and the Y-Balance Test (14, 15). As the mSEBT is a reliable clinical tool to assess dynamic postural control, a recent review with practical guidelines suggested to use this short version instead of the 8-directions SEBT (13). All of these tests require little equipment and are easy-to-use in clinical settings (16).

Even if these tests seem promising regarding the assessment of sprained ankles, both of them have mainly been studied in healthy or chronic ankle instability populations (12, 14, 15, 17). Therefore, the impact of acute pain on reach distance and motor performance remains unknown. If the presence of pain interferes with ankle motor control, it could significantly reduce mSEBT reach distance and adversely affect score interpretation. Hodges and Tucker suggested that acute pain can cause changes in mechanical behaviour (18). These changes could increase muscle stiffness and induce a redistribution of load on joints or affect the direction of force vectors during movement. Such changes could therefore affect performance during the mSEBT.

Moreover, studying the effect of pain on motor control is of great interest as musculoskeletal pain is a major reason for consultation in primary care (19) and can be associated with reduced function (20). However, studying groups with musculoskeletal pain can be very challenging due to high rates of participants' exclusion and to the difficulty to predict how painful a given task will be for each individual (21). Therefore, recruiting healthy participants could avoid these limitations, and allow assessment of the impact of pain on motor control under controlled conditions.

The main objective was to assess the effect of an acute electrical nociceptive pain at the ankle on reach distances during the mSEBT. To do so, participants were divided in three sub-groups (no pain, non-nociceptive electrical stimulation and nociceptive electrical stimulation) to complete two mSEBT and compare their reach distance between the two assessments (the first mSEBT is performed without stimulation for all groups). We hypothesized that if pain has a specific impact on motor control, only participants in the painful group would show a significant reduction for reach distances during the second mSEBT. Therefore, this hypothesis is related to the fact that pain, and not the electrical stimulation, could alter the mSEBT performance.

2. Material and methods

2.1. Participants

Sample size was calculated using G*Power 3.1.9.6 and based on a previous study to determine the optimal number of participants per group (22). A convenience sample of 48 participants was recruited from Université Laval student population. All included participants had to be (1) unaware of the research hypothesis, (2) be aged between 18 and 35 years old and (3) be free of any self-reported pain on the day of the experiment. Participants also had to (4) be free from any lower limb injury in the last 6 months, (5) be able to tolerate an experimental pain of 4/10 on the visual analog scale (VAS) for the duration of assessment and (6) be free of any movement limitation at the lower limb or any neurological impairment that could have affected task performance. Participants were excluded if they scored 71/80 or lower on the Lower Extremity Functional Scale (LEFS), a self-reported questionnaire used to assess lower limb function. The cut-off score of 71/80 was selected regarding its minimum detectable change (MDC) (23). All participants read and signed a consent form describing the experimental procedure and their involvement in the study. This protocol was approved by the local ethics review board (CIUSSS-CN, #2010-212). The experimental procedures were in accordance with the Declaration of Helsinki.

2.2. Modified star excursion balance test

The modified Star Excursion Balance Test is a simplified version of the SEBT. Measuring tapes are placed on the floor in a Y-shaped form and participants have to stand on one foot (the one assessed) in the middle of the Y (**Figure 1**). They are asked to reach as far as they can on the measuring tape while maintaining balance, with their

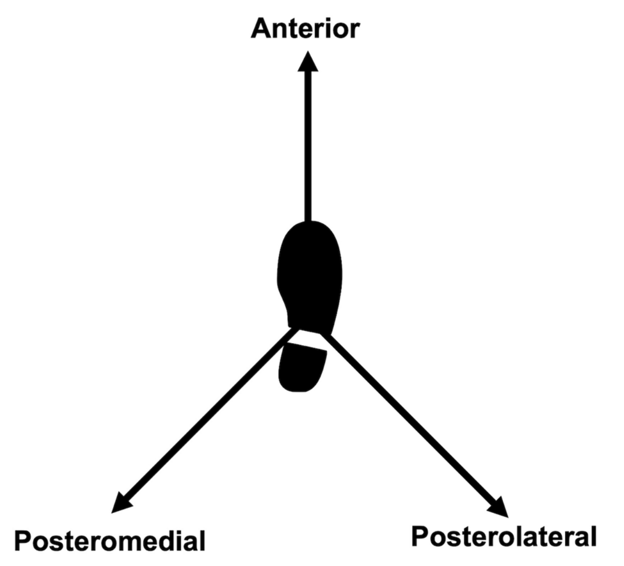

FIGURE 1
Representation of the *modified star excursion balance test* for the right weight bearing limb.

hands on their hips and the stance foot remaining flat on the ground. For a trial to be accepted, participants need to execute a controlled excursion on the Y-shaped form and lightly touch the ground with the tip of their foot as far as they can (24). If they lose balance or step on the measuring tape, they must repeat the trial.

2.3. General protocol

Participants were recruited for a laboratory session that lasted 45 min. Upon arrival, they read the general protocol and completed the LEFS and the Waterloo Footedness Questionnaire (WFQ) to assess foot preference (25). Then, a physiotherapist from the research team measured participants leg length and explained to the participants how to execute the mSEBT. They were asked to practice the mSEBT four times (26) while receiving verbal feedback from the physiotherapist to standardize the mSEBT execution.

Immediately following the practice period, participants had to complete the first mSEBT (mSEBT1). Since none of the participants had sprained ankles, all mSEBT were assessed while standing on the dominant limb based on the WFQ. All participants had to reach as far as they could on the measuring tapes. Reach distance was assessed a minimum of two times for each direction: Anterior (Ant), Posterolateral (PL) and Posteromedial (PM). Since maximum excursion distances values have usually achieved stability within the first 4 practice trials (26), participants were asked to complete each reach distance assessment twice. If the difference between two reach distance measurements was greater than the MDC (i.e., 6.46 cm for Ant, 9.28 cm for PL and 7.55 cm for PM) (8), a third and final attempt was made for this direction. The two closest values were kept for analysis.

After mSEBT1, participants had to remain seated for 15 min. During this break, they were randomly assigned to one of the three following groups: (1) *Control group*, in which participants completed a second mSEBT without any electrical stimulation; (2) *Painless stimulation group*, where participants completed a second mSEBT with a non-nociceptive electrical stimulation at the ankle; and (3) *Painful stimulation group*, in which participants would receive a nociceptive electrical stimulation at the ankle.

For the participants in Painless and Painful groups, during the 15-minutes break, stimulation electrodes were placed on the right lateral malleolus and at the distal end of the fibula of the dominant limb and the intensity of the electrical stimulation was calibrated; thereafter they were asked to complete mSEBT2 with the electrical stimulation. Participants in the Control group were asked to complete a second mSEBT (mSEBT2) following the break without any difference from mSEBT1.

2.4. Electrical stimulation

Two electrical stimulators (s-88, Grass Instruments, Quincy, MA, USA) were used to generate trains of 5 pulses at 300 Hz (pulse width 500µs) delivered through a Digitimer DS7A stimulator (Hertfordshire, United Kingdom) to an anode and a cathode placed two centimeters apart longitudinally over the right lateral malleolus and fibula. The electrodes placement was adjusted for each participant in a way that the pain would be local around the lateral malleolus (i.e., not causing radiating pain). Stimulation was triggered by a foot switch located under the dominant heel and was therefore present during each attempt. For the Painless group, increases in steps of 5 mA were used to individually adjust the stimulus intensity until the perception threshold (i.e., the lowest intensity at which each participant could feel the electrical stimulation) was reached. This 5 mA increment was delivered through a constant current unit in order to standardize the stimulus intensity increment, regardless of skin type or electrode quality. Final stimulus intensity was set at 1.2 times the threshold. For the Painful group, the same increases in steps of 5 mA were used until a pain level of 4/10 on the VAS was reached. For both groups, the intensity remained constant throughout the experiment. For more information regarding this experimental MSK-like pain protocol, see Bertrand-Charette et al. (27).

2.5. Recordings and data analysis

The physiotherapist assessing the mSEBT stood next to the participant during each attempt and noted the reach distance for each direction. Data were recorded for the raw score in centimeters and then normalized according to leg length, where the raw score was divided by the leg length and multiplied by 100 (28).

2.6. Statistics

First, to look at the overall distribution of data and guide the selection of statistical analysis, a violin plot (**Figure 2**) was built with packages ggplot2 (version 3.4.0, 2022-11-04), gridExtra (version 2.3, 2017-09-09) and the function GeomSplitViolin (https://github.com/iholzleitner/facefuns/blob/main/R/geom_split_violin.R) from the R statistical software (version 4.2.2, 2022-10-31). Then using IBM SPSS Statistics 29.0.0.0 (Armonk, NY), a repeated measures ANOVA designed for Gamma distributions (29) (GEE, generalized estimating equations) was used with Holm's sequential Bonferroni correction to compare normalized reach distances between two assessments, three directions and all groups. Some default parameters were changed as followed: DISTRIBUTION = GAMMA, LINK = LOG, and CORRTYPE = UNSTRUCTURED. The three independent variables were Group (between-subjects: Control, Painless and Painful), Time (within-subjects: mSEBT1 and mSEBT2) and Direction (within-subjects factor: Ant, PL and PM). Inherent pairwise comparisons to GEE model with Holm's Sequential Bonferroni were performed as *post-hoc* in the presence of significant GEE results. Significance level was set at 0.05.

3. Results

3.1. Sample size and participants' characteristics

Following sample size calculation, a minimum of 15 participants per group was required to obtain statistical power of 0.95. Thus,

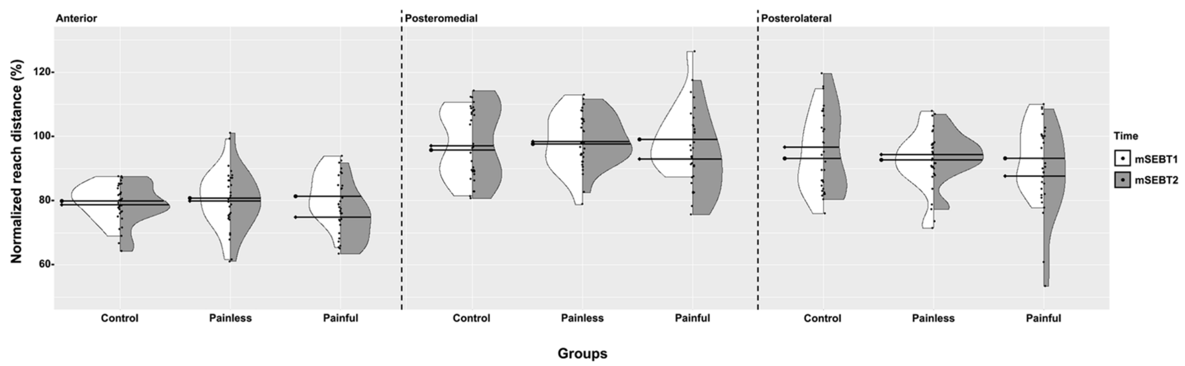

FIGURE 2
Violin plot representing mSEBT1 and mSEBT2 data. The distribution of mSEBT1 (white) and mSEBT2 (grey) data are presented for each group and for each direction. The thick black lines represent the median for each dataset (●- for mSEBT1 and ◆- for mSEBT2). Each dot represents the normalized reach distance for a participant.

forty-eight participants between the age of 19 and 34 years old were recruited for this experiment. Participants' characteristics can be found in **Table 1**.

3.2. Stimulus intensity during the mSEBT2

Participants in the Painful and Painless groups received electrical stimulation during the second mSEBT. Painful group intensity was 9.8 ± 2.4 mA while Painless group intensity was 2.1 ± 1.1 mA.

3.3. Effect of pain on the normalized reach distances

Following the visual inspection of **Figure 2**, the GEE ANOVA was selected as it reported a far better goodness-of-fit statistic when using a Gamma distribution (log link; QICC = 39.39) compared to a normal distribution (identity link; QICC = 27104.50).

TABLE 1 Participants' characteristics.

Characteristic	Control	Painless	Painful	*P*
n	16	16	16	n.s.
Age	27.5 ± 4.2	26.2 ± 4.4	26.8 ± 3.2	n.s.
Sex	8 M; 8F	8 M; 8F	8 M; 8F	n.s.
Footedness	15 R; 1 L	12 R; 4 L	16 R; 0 L	n.s.
Height (cm)	168.9 ± 11.5	174.3 ± 8.2	173.6 ± 8.9	n.s.
Leg length (cm)	91.9 ± 7.4	94.9 ± 5.6	95.4 ± 5.6	n.s.
Weight (kg)	69.1 ± 14.3	73.8 ± 10.8	71.2 ± 9.7	n.s.
LEFS score (/80)	78.3 ± 2.6	78.2 ± 2.4	78.8 ± 1.5	n.s.
Stimulation intensity (mA)	0	2.1 ± 1.1	9.8 ± 2.4	n.a.
VAS score during mSEBT2 (/10)	0	0	4	n.a.

F, female; L, left; M, male; R, right; n.a., not applicable; n.s., not significant VAS, Visual Analog Scale.

Therefore, GEE analysis (see **Table 2**) was applied on the normalized reach distances, that is the raw score divided by the leg length and multiplied by 100 (28).

Group x Time interaction ($p = .000011$), and *post-hoc* tests, indicated that a significant difference between times (mSEBT1 and mSEBT2) happened strictly within the Painful group. No other statistically significant changes exist across groups when comparing within or between Control and Painless groups across mSEBT1 and mSEBT2 (see **Supplementary File 1** for specific results). Therefore, all groups had similar reach distances at mSEBT1 ($p > .05$) and at mSEBT2 for Control and Painless ($p > .05$). On the contrary, regardless of direction, the Painful group showed a statistically significant change between mSEBT1 and mSEBT2 (7.03 ± 1.46% [2.76, 11.30], $p = .00002$).

As it was previously reported that Ant, PL and PM reach distances are affected differently by various factors (30), further *post-hoc* tests were performed for each direction to better understand this effect (**Table 2**). Moreover, effect sizes were examined as mean absolute differences with 95% confidence intervals reported between brackets. The *post-hoc* inherent pairwise comparisons reported no significant difference at all among the Control and Painless groups for all directions ($p > .05$) when comparing mSEBT1 and mSEBT2, while reach distance significantly decreased for the Painful group in the anterior (−7.06% [1.68, 12.43], $p = .00048$) and the posteromedial (−6.53% [0.80, 12.26], $p = .0075$) directions. The posterolateral distance showed a decrease of 7.34% which is consistent with the decrease seen in the two other directions although this difference is not statistically significant ($p = .10$).

4. Discussion

This study aimed to assess the effect of acute electrical nociceptive stimulation at the ankle on mSEBT reach distances. All groups performed two mSEBT separated by a 15-minutes break. However, only the Painful group showed significant reduction in reach distances during the second mSEBT. Our results suggest that acute pain could alter lower limb motor control.

TABLE 2 *Post-hoc* test results for normalized reach distance.

Group	Direction	mSEBT1 [mean ± SD]	mSEBT2 [mean ± SD]	Sequential Bonferroni Sig.	95% Wald Confidence Interval for Difference		Mean difference (%) [mSEBT1–mSEBT2 ± SD]
					Lower	Upper	
Control	Ant	79.64 ± 5.30	78.59 ± 6.81	$p > .05$	−1.32	3.43	1.05 ± 0.96
	PL	93.98 ± 11.19	95.72 ± 13.20	$p > .05$	−4.65	1.18	−1.74 ± 1.03
	PM	97.24 ± 10.19	98.05 ± 11.66	$p > .05$	−2.53	0.91	−0.81 ± 0.68
Painless	Ant	80.50 ± 9.07	80.06 ± 9.54	$p > .05$	−0.81	1.69	0.44 ± 0.54
	PL	92.03 ± 10.06	93.72 ± 7.93	$p > .05$	−4.29	0.92	−1.69 ± 0.86
	PM	98.13 ± 9.29	98.97 ± 7.63	$p > .05$	−2.62	0.93	−0.84 ± 0.70
Painful	**Ant**	**81.92 ± 7.79**	**74.86 ± 8.27**	**$p = .00048$**	**1.68**	**12.43**	**7.06 ± 1.54**
	PL	94.15 ± 10.07	86.77 ± 14.85	$p > .05$	−0.46	15.21	7.37 ± 2.30
	PM	**99.83 ± 10.47**	**93.30 ± 12.17**	**$p = .0075$**	**0.80**	**12.26**	**6.53 ± 1.66**

Bold values represent statistically significant changes.

4.1. Effect of pain on the reach distances

Participants in the Painful group showed a significant decrease in reach distance for two out of three mSEBT directions in the presence of an acute electrical nociceptive stimulation. This reduction ranged from 6.53% in PM to 7.06% in Ant for the normalized scores. These results are similar to previous studies comparing chronic ankle instability to a control group (17, 24, 31–33), where all participants with chronic ankle instability showed significant decrease in reach distances compared to controls. Only one study looked at the impact of acute ankle sprains on the reach distance (34). Similar to the chronic ankle instability studies (17, 24, 31–33), they noted a decrease in reach distance when comparing the sprained ankle group to the control group. However, there was no information regarding pain intensity from the participants in the acute ankle sprain groups making it hard to conclude on the impact of pain on the mSEBT reach distance. The presence of experimental acute pain in the present study caused a decrease in reach distance similar to what is seen with acute and chronic sprained ankles. This suggests that pain could negatively influence lower limb motor control even in the absence of mechanical limitation and that Ant and PM directions might be more affected by acute experimental pain than PL. However, it is important to note that the decrease seen in PL, even though not significant, is similar to the Ant and PM directions. Therefore, by recruiting more participants, PL could eventually show the same significant reach distance decrease. Moreover, a previous study showed that following ankle sprains, alteration in ankle motor control is not only the result of a peripheral deficit, but likely to be second to a reorganization of central motor commands, resulting in bilateral deficits during the SEBT (33). Therefore, pain and ligaments structural integrity both have the potential to interfere with motor control and general stability in sprained ankle participants.

4.1.1. Clinical relevance

Another important finding in the present study is that the reach distance reduction caused by pain is greater than the minimal detectable change (MDC) of the SEBT for ANT direction. For example, the normalized score MDC for ANT has been reported to be 5.87% (35) while our results show a 7.06 ± 1.54% reduction in reach distance. The MDC is an estimate of the smallest change that falls outside the measurement error in the score, and it is based on the standard error of the mean (8, 36). Therefore, the mean 7.06 ± 1.54% shown in our results suggests that some participants had a reduction in reach distance with pain that was greater than the measurement error. It is also important to mention that the 95% confidence intervals were quite large, ranging from 1.68 to 12.43%. This supports the fact that pain is a personal experience (37) and that it might affect motor control differently, even across participants showing similar personal characteristics. These results, specific to the ANT direction, could suggest that this direction is the most affected by pain, in terms of motor control. As a matter of fact, a decreased performance in this direction has been shown to be related to an increased lower limb injury risk (38, 39). This direction is also highly affected by ankle dorsiflexion angle (40), a parameter shown to be reduced following chronic ankle instability and described as a predisposing factor for ankle injuries (41).

Finally, regarding the Painless and Control group, no significant changes were found in all three directions. This further support the hypothesis that pain and not just an electrical stimulation or distraction can alter motor control during a functional task. Moreover, none of these groups reach distance increased following the first mSEBT. This means that there was no learning effect throughout the study that could have affected the second execution of the test or the results.

4.2. Interaction between motor control and pain

Motor control is defined as the ability to regulate or direct the mechanisms essentials to movement (11). For proper regulation, timely integration of sensory information with movement planning and execution (i.e., sensorimotor integration) is necessary (8). The fact that spinothalamic projections to the motor cortex have been shown in humans (42) suggest that nociception should be considered both as a sensory input and also as a potential contributor to motor control. This contribution could either be beneficial or detrimental to performance during a functional task. In the current study, nociceptive inputs were detrimental to motor control during the modified Star Excursion Balance Test (a valid test used to assess motor control) by reducing reach distance in the Painful group. This interference of nociception on motor control is supported by neurophysiological studies [see Bank et al. (43) and Rohel et al. (44) for systematic reviews]. For example, M1 and S1 have been shown to exhibit decreased excitability in the presence of acute experimental pain (43, 45, 46). In addition, the Motor adaptation to pain model of Hodges and Tucker suggests that changes in mechanical behavior resulting from altered motor unit recruitment could be present around joints when pain is present (18). These changes could modify muscle stiffness and/or motor unit recruitment, here again affecting motor control. Finally, a recent study assessing proprioceptive acuity while walking demonstrated that pain can also interfere with sensorimotor integration during functional tasks (22). These studies, combined with the findings from the current study, demonstrate that pain interferes with sensorimotor integration and movement production, resulting in impaired motor control.

4.3. Strengths and limitations of the study

This study has some limitations. First, participants in all three groups were relatively young adult, which might limit the generalizability of the results. Another limitation is the absence of kinematic variables that could have added more detailed information on lower limb displacement during the mSEBT. Finally, the number of participants in each group was relatively small, resulting in large 95% confidence intervals for the mean absolute differences in reach distances.

This study also has several strengths. It is the first study to look at the effect of acute experimental pain on lower limb motor control (as assessed by the mSEBT). The presence of a group receiving non-nociceptive electrical stimulation allowed us to conclude that it is actually pain and not simply the electrical stimulation that specifically caused the modification in lower limb motor control. Finally, the use of an electrical nociceptive stimulation that caused a focused, acute and easily adjustable pain made it possible to control this pain intensity across participants in the Painful group.

5. Conclusion

Our results show that acute ankle experimental pain causes a reduction in mSEBT Ant and PM reach distances. This suggests that acute pain has the potential to interfere with lower limb motor control. Clinically, if the presence of pain interferes with ankle motor control, it could mean that the interpretation of the mSEBT reach distance should take into account the presence of pain, as it can significantly reduce the participant's ability to reach further. Further studies should include patients with acute painful ankle sprains to compare their results with the nociceptive electrical stimulation group to assess the effect of MSK pain on ankle motor control.

Data availability statement

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by Comité d'éthique de la recherche sectoriel en réadaptation et intégration sociale, Centre intégré universitaire de santé et de services sociaux (CIUSSS) de la Capitale-Nationale (CIUSSS-CN, #2010-212). The patients/participants provided their written informed consent to participate in this study.

Author contributions

MB-C, J-SR and LB contributed to study conception and design. MB-C conducted data collection. MB-C and LB performed data validation and analysis. MB-C wrote the draft of the paper and prepared figures. All authors provided substantive feedback on the paper. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2023.1082240/full#supplementary-material.

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Beyond physiology: Acute effects of side-alternating whole-body vibration on well-being, flexibility, balance, and cognition using a light and portable platform A randomized controlled trial

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A good body-balance helps to prevent slips, trips and falls. New body-balance interventions must be explored, because effective methods to implement daily training are sparse. The purpose of the current study was to investigate acute effects of side-alternating whole-body vibration (SS-WBV) training on musculoskeletal well-being, flexibility, body balance, and cognition. In this randomized controlled trial, participants were randomly allocated into a verum (8.5 Hz, SS-WBV, $N = 28$) or sham (6 Hz, SS-WBV, $N = 27$) condition. The training consisted of three SS-WBV series that lasted one-minute each with two one-minute breaks in between. During the SS-WBV series, participants stood in the middle of the platform with slightly bent knees. During the breaks in between, participants could loosen up. Flexibility (modified fingertip-to-floor method), balance (modified Star Excursion Balance Test), and cognitive interference (Stroop Color Word Test) were tested before and after the exercise. Also, musculoskeletal well-being, muscle relaxation, sense of flexibility, sense of balance, and surefootedness were assessed in a questionnaire before and after the exercise. Musculoskeletal well-being was significantly increased only after verum. Also, muscle relaxation was significantly higher only after verum. The Flexibility-Test showed significant improvement after both conditions. Accordingly, sense of flexibility was significantly increased after both conditions. The Balance-Test showed significant improvement after verum, and after sham. Accordingly, increased sense of balance was significant after both conditions. However, surefootedness was significantly higher only after verum. The Stroop-Test showed significant improvement only after verum. The current study shows that one SS-WBV training session increases musculoskeletal well-being, flexibility, body balance and cognition. The abundance of improvements on a light and portable platform has great influence on the practicability of training in daily life, aiming to prevent slip trips and falls at work.

KEYWORDS
whole-body vibration (WBV), musculoskeletal, flexibility, balance, cognition, inhibition, stroop-color-word interference task, SLIP trip and fall accidents

Introduction

Slips, trips and falls (STF) are the most frequent accidents in Switzerland, causing 27.7% of all accidents, i.e., about 70,000 workers, in 2020 (1). Also, STF are the most expensive accidents and yielded 41% of all accident expenditure for the years 2014 to 2018 combined. In order to be able to reduce STF incidents, risk factors must be identified (2).

One important risk factor of falls is a weak balance (3), which is related to muscle weakness (4). However, to predict fallers is difficult, because there are several risk factors, which include motor, sensory, and cognitive processes (3). Due to loss of balance being a possible influence of individual frailties on STF (5), balance trainings are recommended to reduce STF (6, 7). Most tested balance trainings often are time-consuming, need much advise, put other regulatory efforts like change of clothes or place which increase regulatory demands at work and therefore low compliance and drop out often occurs (8, 9). The training goal with respect to prevention is that the training is short and easy to administer but also easy to adjust to individual condition and there is no need for change of clothes, shoes, or location (10). In addition, there should be a benefit from each single training session that is noticeable. Thereby, it would be an advantage, when the benefit of a single training is not only improved body balance but includes other improvements like improved mental functions as well. Multiple benefits make it more likely that the training is accepted and becomes a routine behaviour. Also, to increase long-term adherence, it is important to build a routine around a person's lifestyle (11).

In the current study, whole-body vibration (WBV) training is introduced as an exercise-based health-intervention to improve balance with the aim to reduce STF not only in terms of improving motor and sensory, but also cognitive processes.

Healthy and unhealthy forms of whole-body vibration

Long lasting vibrations are biomechanical risk factors that contribute to the development of musculoskeletal pain (12). Other mentioned biomechanical risk factors are heavy load lifting, bending and twisting and remaining in a static position over longer time (13). Vibration exposure from driving vehicles or from vibrating, hammering or rotating work equipment may lead to musculoskeletal and neurological disorders, depending on strength, frequency, duration of action, working method and body posture (14, 15). For example, vibration experienced by construction workers handling compressed air hammers or truck drivers during long-term journeys can cause vascular, neurological and musculoskeletal problems, as well as disturbances of the lumbar spine and the associated nervous system (16).

However, a large number of studies have shown that shorter exposure on vibration can also have a prophylactic effect on musculoskeletal discomfort when range of vibration frequency, amplitude and duration are properly dosed. Thus, in addition to a reduction of musculoskeletal disorders, WBV training can also promote improvements in sensorimotor and muscular performance, balance, functional mobility, bone mineral density, maximum and rapid force, stretch reflexes, and speed of movement (17–25).

WBV exercises are easily applied (10). As the exercise is not exhausting, users usually do not sweat during training sessions. WBV is easily adapted to individual level of body balance and fitness. Hence, users do not have to change clothes or shoes, or take a shower afterwards, which could be important in occupational settings or in healthy young adults, where users do not want to waste time on an intense worksite activity training. High training durations often result in a lack of participation and compliance rate (8, 9). According to worksite training studies, the duration of one WBV training session is about 10 min (26, 27), which is half the time participants usually have to invest in worksite activity trainings (28).

After brief instructions concerning the correct body posture and the handling of the vibration platform, participants can start WBV exercises, which have proven to gain high compliance rates (26, 29). A three-month WBV-intervention with employees suffering from chronic low-back pain, revealed a compliance rate of 81.1%, with two to three recommended trainings per week (29). A four-week WBV-intervention study with office-workers of a Swiss federal department even revealed a training attendance of 129%, therefore more than the instructed three trainings per week (26).

Different forms of whole-body vibration training

In their systematic review, Oliveira and colleagues (30) found adverse events in only 55 of 1,833 volunteers, who mentioned experiencing for example back pain, pain in their legs, or dizzy sensations. With only 3% adverse events, WBV training is therefore considered to be relatively safe (30). Rogan and colleagues (31) have differed three types of WBV. Sinusoidal vertical WBV (SV-WBV) and sinusoidal side-alternating WBV (SS-WBV), which use a single vibrating platform, and stochastic resonance WBV (SR-WBV), which functions with two independent powered platforms, which can be comparable to skis.

Vibration frequencies among sinusoidal WBV are constant, whereas SR-WBV works with unpredictable random frequencies forcing the human body to constantly adapt its neural and muscular reactions (32). To the best of our knowledge, SR-WBV was originally developed to increase performance of professional ski athletes. Nowadays it is used in different sports as injury prevention, but also in therapies with Parkinson patients (33), stroke patients (34), frail elderly (32), or patients with chronic low back pain (35). However, devices working with SR-WBV are big, heavy and expensive and thus rather used in physiotherapy or in fitness centers than at home.

In the current study, SS-WBV is applied by using a light and portable platform, which seems to be ideal as a training device whether working from home or onsite. Although, SS-WBV seems to have a higher effect than other forms of WBV on bone mineral density (30), in terms of load, Rohlmann and colleagues (36) showed that the maximum load on the vertebral body was lower in SS-WBV (15%) than in SV-WBV (27%). This might be due to the fact, that in comparison to SV-WBV where both legs move up and

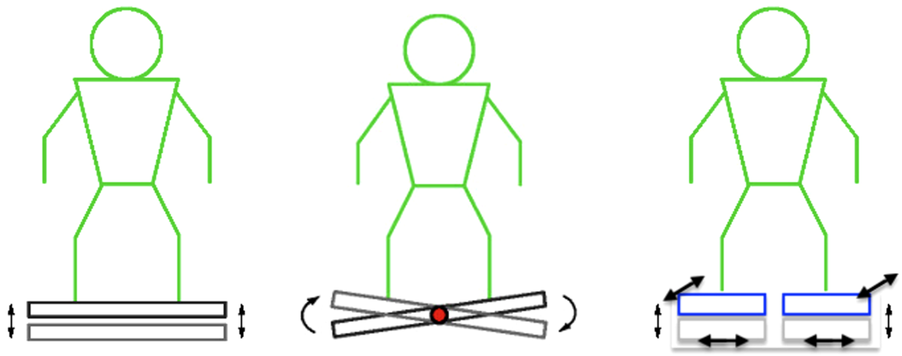

FIGURE 1
Different types of vibrating platforms. Whole-body vibration (WBV) types from left to right: sinusoidal vertical (SV-WBV), sinusoidal side-alternating (SS-WBV), and stochastic resonance (SR-WBV); source: mediplate.ch; SR-WBV customized by the author.

down at the same time, in SS-WBV, the oscillations take place around a pivot at the center of the platform. Due to this, users have to alternate vibrations between both sides, i.e., when the right foot moves up, the left foot moves down, and vice versa (30, 37). The different types of vibrating platforms are illustrated in **Figure 1**.

WBV training has proven to be a safe and useful way to improve performance in athletes among different sports (38). However, as elite athletes already have a high level of performance, WBV training often leads to bivalent results, because it might be too unspecific to improve sport-specific strength, flexibility or balance (39, 40). Not only have recent studies revealed that WBV training is especially beneficial to improve stability or functional mobility in the elderly population (41), in patients with low-back pain (42) or in stroke patients (43), promising effects, such as increased balance and musculoskeletal well-being were also found at the workplace, e.g., in office workers who spend much time in sitting position (26). Recently, even individuals affected with COVID-19 who performed WBV training have exhibited improvement in inflammatory status and an overall improvement in quality of life. Moreover, a reduction of time in intensive care units in severely affected patients was also identified (44).

The current study aims to add knowledge on SS-WBV training effects by use of an experimental design. Experimental evidence for SS-WBV training effects is an important first step before an examination of this portable platform as training device for work from home and onsite occurs in future studies. Therefore, the focus of the current study is on acute musculoskeletal and cognitive effects of SS-WBV in laboratory, unprecedentedly using a light and portable platform with young and healthy participants. Although SS-WBV effects would be expected to be stronger in an older and unhealthy sample, effects are also expected to be meaningful in young and healthy individuals.

Another goal of the current study is to be able to observe side-effects. Health risks increase simultaneously as vibration intensity and exposure increase. However, according to Seidel et al. (1986), vibrations under 20 Hz are safe (45). In the current study, frequencies under 10 Hz are applied, thus side effects are not expected. If no side-effects are observed, future studies might include older people or frail individuals as well.

Cognitive interference and its relations with physical parameters

Body balance depends on musculoskeletal and cognitive function. Together with attention, cognitive flexibility and decision making, inhibitory control belongs to the executive functions (46). Executive functions are located in the prefrontal cortex and are responsible for higher order cognitive abilities, e.g., volitational control over goal-directed behavior (47–50). Thus, goal-directed behavior may be volitionally achieved by deliberately suppressing dominant, automatic responses or impulsive reactions (49, 51). For example, when we walk in the park our automatic tendency would be to place one foot in front of the other. However, when suddenly facing a slippery or unstable surface, inhibitory control helps us to stop this automatic behavioral tendency, which must quickly be modified (52).

Two subcomponents of inhibitory control are motor response inhibition, i.e., the process of revoking an impulsive reaction, and cognitive interference inhibition, i.e., the ability to withstand stimuli related interference of the external environment (53, 54). The latter is subject of the Stroop Test and is subsequently referred to as "cognitive interference".

Due to the incongruent occurrence of two stimuli (color and description) in the validated Stroop Test (55), the examinee perceives the occurrence of the stimuli as unwanted, sometimes even disturbing, which are two of the defining characteristics of cognitive interference (56). To enable the required performance, participants must ignore the written name of the color. This allows them to name the ink color of the word, which is a goal-directed behavior that requires a little more processing time. Hence, a resulting correlation with poorer performance seems obvious (56).

According to Bolton and Richardson (2022), inhibitory control has proven to be a significant and unique factor in fall prevention (52). Motor training combined with cognitive interference tasks plays an important role, especially for people with Parkinson's disease (46) or older people who participate in fall prevention training (57). These results prompted us to go a step further. Fall-safe older people are more active and safer than their peers, which in turn can lead to a change in physical well-being and not only includes physical activity and balance, but flexibility as well (58).

As recently demonstrated, vibration can improve balance in older people (59), as well as in individuals with metabolic syndrome (60) preventing falls and injuries. Thus, it could be that vibration contributes to improved surefootedness. Faes et al. (2018) were able to show that WBV improved surefootedness - and that WBV has a positive effect on balance in addition to surefootedness in healthy individuals (26).

Regular training is especially important with regard to balance, which is an important part of physical well-being, daily mobility and therefore general ability to function in everyday life (61). This is consistent with improved neuromuscular control leading to a better postural stability achieved through WBV training (62). Furthermore, good postural stability has a positive influence on balance (58).

Specifically with regard to the young and healthy participants in the current study, McClain and Shallen (2015) demonstrated that WBV can improve participants fitness, thus also balance, better than static training. In another study conducted by Despina et al. (2020), WBV training resulted in superior short-term performance improvements in flexibility, strength and balance compared to an equivalent exercise without vibration (63) and thus confirmed similar results from Ritzman et al. (2014) (64). Exercise and training programs that include WBV can therefore provide additional benefits for young and well-trained adults.

The current study aims to find new insights in WBV and their effect on musculoskeletal well-being, muscle relaxation, sense of flexibility, sense of balance, and surefootedness. Specifically, based on previous research as stated above, musculoskeletal well-being, flexibility, and balance should be increased and cognitive interference decreased after one training session of SS-WBV. These results are hypothesized to be found only in the experimental group (8.5 Hz) and not in the control group (6 Hz):

Musculoskeletal well-being and muscle relaxation assessed with questionnaire is expected to be increased after one training session of SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz (H1). Also, flexibility assessed through the modified fingertip-to-floor method (mFTF) is expected to be increased after one training session of SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz (H2). Furthermore, balance measured with the modified Star Excursion Balance Test (mSEBT) is expected to be increased after one training session of SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz (H3). Lastly, cognitive interference measured with the Stroop Color Word Test is expected to be decreased after one training session of SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz (H4).

Materials and methods

Ethics

The study was performed in consensus with all requirements defined by the Swiss Society of Psychology and was conducted with the understanding and the consent of the human subject. The Ethical Committee of the responsible University faculty (University of Bern) has approved the study (Nr.: 2019-07-00005).

Participants

Number of participants was calculated using G-power software. A moderate effect size was chosen as a standard in this calculation (65). The required sample size for each exercising condition – verum and sham - was 28 participants, expecting a moderate effect size ($d = 0.5$) for the t-test analysis between two dependent means and a requirement of 90% power.

Participants with one or more of the following criteria were excluded: Being pregnant, having osteosynthesis material (such as implants or screws) in the body, musculoskeletal disorders, joint problems (especially regarding the knee, hip, and back), herniated discs, rheumatism (such as spondylitis, gout, osteoporosis, osteoarthritis), cardiovascular complaints, disorders related to the sense of balance (such as hearing loss). Also, participants were advised to attend the study in a rested state and must not have had any intensive workout within the previous 24 h, because of musculoskeletal and cognitive effects. In order to attend the Stroop Test (66), participants must also not suffer from red-green color blindness or take medication known to affect the central nervous system.

A number of 55 students and acquaintances signed up for the study. No participants had to be excluded before, during or after the experiment. Body mass index was calculated as a participants weight in kilograms divided by height in meters squared. Students who participated were reimbursed with one of 15 mandatory participant-hours by the associated university. Acquaintances were thanked with sweets after the experiment.

Vibrating Platform

Two vibration platforms named MediPlate® (Dormena GmbH, Liestal, Switzerland) were used in the current study. They reach frequencies between 6 and 13 Hz of ball-bearing side-alternating (rocking) vibrations. The MediPlate® represents a transportable vibration platform as it weighs only 15.5 kg and is rather small (length: 77 cm, width: 44 cm, height: 12.5 cm). Amplitude is between 2 mm and 8 mm depending how participants place their feet on the platform. In the current study, participants exercised with an amplitude of about 5 mm.

The verum condition was set at a frequency of 8.5 Hz (Level 20). It is experienced as slightly higher than the minimal stimulation parameter of 6 Hz (Level 1), which was used as sham condition. Acceleration forces – calculated as $f(\max) = \text{amplitude} * (2\pi * \text{frequency})^2$–were 12.8 m/s^2 (1.3 g) in the verum condition and 5.3 m/s^2 (0.5 g) in the sham condition. Thus, forces of the verum condition on the body were lower than walking, which reaches between 2.7 g and 3.7 g (67).

Since there were no studies on the MediPlate® vibration platform so far, our decisions concerning chosen frequencies rely on experience with various vibration training studies (10) combined with recommendations from the designers of MediPlate®. A blank control group without any vibration was not carried out to ensure that participants were unaware of their group-allocation.

Musculoskeletal well-being and muscle relaxation assessed with questionnaire

Musculoskeletal well-being and muscle relaxation were assessed with a short version of the self-administered questionnaire of Burger et al. (2012) before and after the exercise (3). The questions started with the lead-in phrase, “How do you rate your personal sensations regarding muscles and joints (back, shoulders and neck, legs) at this moment?” and were answered on a 100-point-rating-scale from zero (“not at all comfortable/relaxed”) to 100 (“as comfortable/relaxed as possible”).

Flexibility assessed through the modified fingertip-to-floor method (mFTF)

Flexibility was assessed through the modified fingertip-to-floor method (mFTF). Compared to the original fingertip-to-floor method (FTF), where participants stand on the floor, participants stand on a box when attending the mFTF. This is an advantage as measurements of participants who are able to touch the floor or reach beyond can still be included (68). Participants are asked to bend over as far as possible keeping their legs and arms straight, while the examiners measured the distance to the box. This procedure was repeated three times. Gauvin et al. (69) reported high test-retest ($r = 0.98$), as well as high inter-rater reliability ($r = 0.95$) for the mFTF.

Additionally, sense of flexibility was assessed in one question: “How flexible do you feel at this moment?” (17). Answers could range from 0 being “a lot worse than usual”, to 100 being “much better than usual”, and with 50 being “same as always”.

Balance measured with the modified star excursion balance test (mSEBT)

Balance was measured with the modified Star Excursion Balance Test (mSEBT) (70). In this test, dynamic balance is assessed in the eight directions anterior, anteromedial, medial, posteromedial, posterior, posterolateral, lateral und anterolateral with both high intra-test ($r = 0.84$ to 0.93) and test-retest reliability ($r = 0.89$ to 0.93) (70, 71). Excursion distances were normalized to individual leg length of each participant, in order to exclude effects related to gender, because males were found to have significantly greater excursion distances than females (72). Pozo-Cruz et al. (2011) stated that previous studies found similar balance-test results for the dominant and non-dominant leg (73). Also, in the current study, balance of both legs dominant and non-dominant were measured.

Additionally, sense of balance and surefootedness were assessed with two questions: “How do you rate your personal feelings about your balance at this moment?” and “How sure-footed do you feel at this moment?” (17). Answers could range from 0 being “a lot worse than usual”, to 100 being “much better than usual”, and with 50 being “same as always”.

Cognitive interference measured with the stroop color word test

Cognitive interference was measured with the Stroop Color Word Test (66). In this well-established test (74), participants are given color words that are written in color and are asked to indicate the ink color of the word, thus having to ignore the dominant tendency of reading the word.

In the current study a digital form of the Stroop Color Word Test was applied, using the Inquisit 5 Lab program (Millisecond Software, LLC, Seattle, USA) on a computer. After a test trial, the experimental trial started. It consisted of 84 randomly sampled items with congruent (color word, e.g., “red” and the ink color it is presented in is the same, hence red), incongruent (color word, e.g., “black” and the ink color it is presented in is not the same, e.g., green) and neutral items (colored rectangles in black, red, blue or green). Test duration was approximately 3 min.

Laird et al. (2005) describes cognitive interference to be the difference between incongruent items and a control condition, either congruent, neutral, or non-lexical items (75). Analogous to a previous study on cognitive effects from SR-WBV (76), congruent items are compared to neutral items in the present study. A higher difference between both conditions means higher cognitive interference and thus lower inhibitory control (77).

Because keyboards often differ in latency-time (78), reaction response boxes V1.0 (© immo electronics) were applied instead of keyboards. Four buttons, according to the four presented colors, were placed between participant and computer screen. Participants held index- and middle finger of each hand on the buttons during the test. **Figure 2** shows the Stroop Test on a computer set-up.

Procedure

The experiment was carried out by 2 examiners (MH, SS) in a laboratory room at the University of Bern. While the first examiner guided the participants through the procedure, the second examiner acted as an assistant. These roles were changed regularly to prevent monotony. In order to standardize the procedure, examiners followed a strict case report form (CRF). No more than one participant could attend the experiment, which lasted approximately 50 min.

Before participants arrived, they were randomly allocated to verum (8.5 Hz) or sham (6 Hz) group by flipping a coin. Participants then read through the study-information and signed a consent form to declare their voluntary participation and the possibility to stop the experiment whenever they wanted. Although participants were unaware of their group-allocation or their vibration frequencies, blinding of the examiners was not feasible. The first examiner explained the overall procedure but did not reveal the group allocation. Participants attended the baseline-measurements, starting with the baseline questionnaire, Stroop Color Word Test, flexibility test (mFTF) and balance test (mSEBT).

As in previous studies using an SR-WBV vibrating plate (17, 26), the vibration exercise with the MediPlate® consisted of three series that lasted one-minute each with two one-minute breaks in

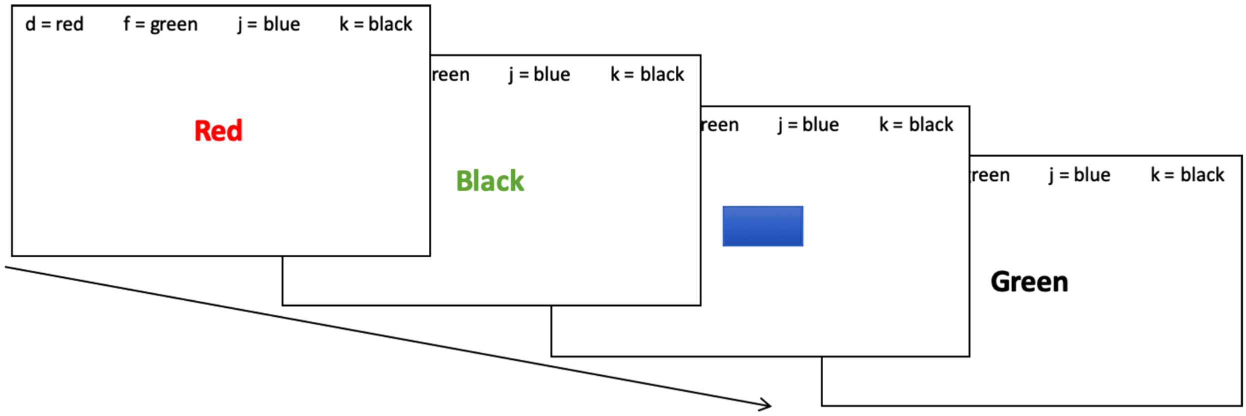

FIGURE 2
Stroop test on a computer set-up. The Stroop Color Word Test was held on a computer. In a test trial, congruent (color word and the color it is presented in are the same), incongruent (color word and the color it is presented in are not the same) and neutral items (colored rectangles in red, green, blue or black) were shown. After a test trial, the 3-minutes experimental trial with 84 randomly sampled items started.

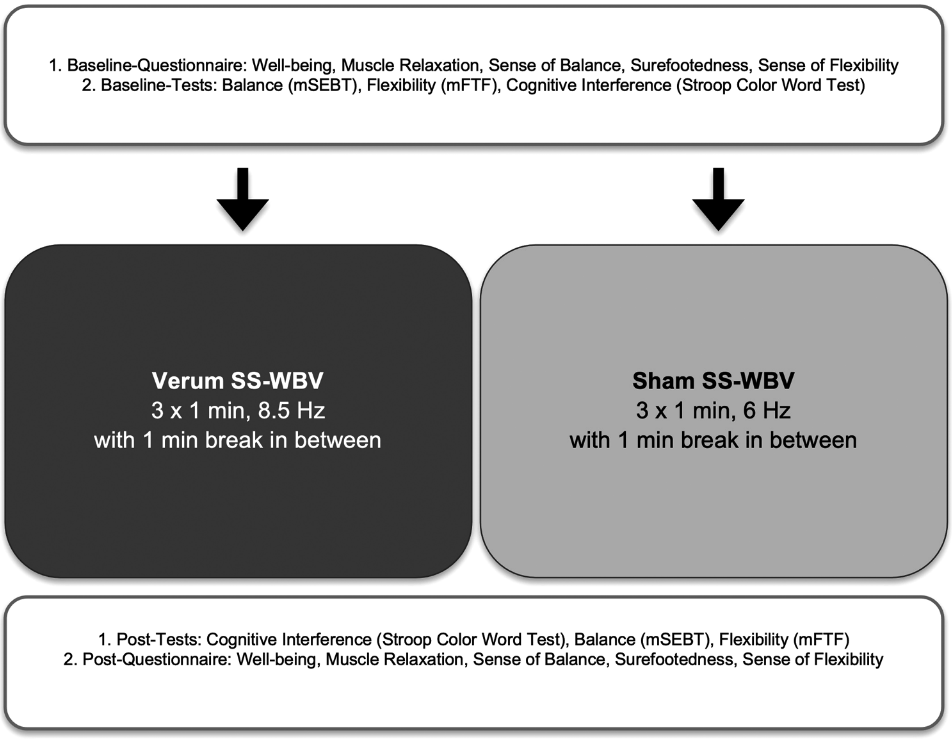

FIGURE 3
Flowchart of the procedure. Baseline-measurements started with the baseline questionnaire and were followed by the Stroop Color Word Test, flexibility test (mFTF) and then balance test (mSEBT); The sinusoidal side-alternating vibration exercise with the MediPlate® consisted of three series that lasted one-minute each with two one-minute breaks in between; Focusing on immediate effects on cognitive interference, participants attended the post-tests (starting with the Stroop Color Word Test, followed by the mSEBT and then mFTF) before the post-questionnaire.

between. Participants were instructed to stand in the middle of the platform facing forward in an upright position with slightly bent knees (i.e., a skiing posture) and with their arms hanging loosely at their sides. In the short break in between, participants could loosen up and prepare for the following series.

Focusing on immediate effects on cognitive interference, participants attended the post-measurements in the following order straight after the exercise: Stroop Color Word Test, flexibility test, balance test and questionnaire. The post-questionnaire was longer as it also included demographical questions. The whole procedure is shown in Figure 3.

Statistical analysis

Musculoskeletal well-being, muscle relaxation, flexibility, balance, and cognitive interference were analyzed in a dependent sample t-test examining differences between baseline and exercising conditions

TABLE 1 Descriptive and inferential statistics.

Variable	Verum SS-WBV 8.5 Hz ($n = 28$)		Sham SS-WBV 6 Hz ($n = 27$)		t	P
	M	SD	M	SD		
Sex (m, f)	6m, 22f		5m, 22f		.27	.792
Age (years)	22.36		23.74		−.81	.423
BMI (kg/m²)	21.23		21.58		−.54	.591
Sport (1 “never”–7 “daily+”)	4.79	.92	5.15	1.17	−1.28	.205
Smoking (yes, no)	4y, 24n		3y, 24n			
BL Well-being	75.64	31.84	80.41	18.20	−.68	.50
BL Muscle Relaxation	73.39	30.62	72.56	19.01	.12	.90
BL Balance-Test (dom. leg)	74.81	10.39	72.41	7.28	.99	.327
BL Balance-Test (non-dom. leg)	73.45	7.03	72.82	5.98	.36	.723
BL Sense of Balance	47.50	6.12	48.96	6.50	−.86	.394
BL Surefootedness	48.86	4.20	51.22	7.07	−1.52	.136
BL Flexibility-Test	3.77	9.69	6.25	11.03	−.89	.379
BL Sense of Flexibility	44.04	8.47	48.48	11.85	−1.61	.114
BL Cognitive Interference (RT control)	145.96	136.77	49.24	439.90	−1.05	.298

Sport: Amount of Sport was measured using a 7-point likert-scale from 1 “never” to 7 “several times a day”; being a smoker was answered with yes (y) or no (n); Baseline (BL); BL Variables: Musculoskeletal well-being and muscle relaxation were assessed on a 100-point-rating-scale from 0 “not at all comfortable” to 100 “as comfortable as you can imagine”; Balance was measured with the modified Star Excursion Balance Test (mSEBT); Sense of Balance and Surefootedness were answered from 0 “a lot worse than usual” to 100 “much better than usual” and with 50 being “same as always”; Flexibility was assessed with the modified Fingertip to Floor Test (mFTF); Sense of Flexibility was answered from 0 “not at all flexible” to 100 “as flexible as I can imagine”; Cognitive Interference was calculated as the difference between reaction time (in milliseconds; ms) in incongruent trials minus the reaction time (ms) in control trials. Higher interference stands for lower inhibitory control; p-values are two-tailed with an α-level set at 5%.

using SPSS (version 25, SPSS, IBM Inc., United States). P-values were two-tailed with an α-level set at 5%. Collected variables were not approximately normally distributed ($p < .05$) as assessed by Shapiro–Wilk Tests. Thus, graphical approaches, skewness and kurtosis were included in the decision, showing all variables to be close to normal. Also, according to Field (79) analysis of the hypotheses can be considered robust against violations of the normal distribution when the group size is equal. Pearson's descriptive statistics for the collected variables are shown in **Table 1**. Results of t-tests for each exercising condition are shown in **Table 2**. Effect sizes are, according to Cohen (80) described as *small* ($d = 0.2$), *medium* ($d = 0.5$), *and large* ($d \geq 0.8$). The formula for the calculation of effect sizes for dependent t-test results is according to Dunlap and colleagues (81):

$$d = t_c\sqrt{\frac{2(1-r)}{n}}$$

Results

Participant characteristics

Fifty-five healthy students and acquaintances (44 female; mean age = 23.04 years, $SD = 6.33$ years; mean height = 170.39 cm, $SD = 9.09$; mean weight=63.09, $SD = 11.29$; mean BMI = 21.40, $SD = 3.03$) took part in the current study. All participants were randomly assigned to verum ($N = 28$) or sham ($N = 27$) condition.

Verum and sham groups did not differ significantly in any demographic characteristics or in baseline variables (**Table 1**).

Higher musculoskeletal well-being and muscle relaxation after verum SS-WBV (H1)

A significant effect on musculoskeletal well-being was found after verum SS-WBV ($t = -2.26$ $p = .032$, $N = 28$), but not after sham SS-WBV ($t = 0.93$, $p = .359$, $N = 27$). Compared to baseline measurement (verum: 75.64 ± 31.84; sham: 80.41 ± 18.20), musculoskeletal well-being increased significantly after verum (88.50 ± 11.54), but not after sham SS-WBV (76.81 ± 26.15). Effect sizes using Cohen's d (75) on musculoskeletal well-being in the verum condition was $d = -0.495$, and in the sham condition $d = 0.151$.

A significant effect on muscle relaxation was found after verum SS-WBV ($t = -2.21$, $p = 0.032$, $N = 28$), but not after sham SS-WBV ($t = -1.16$, $p = .258$, $N = 27$). Compared to baseline measurement (verum: 73.39 ± 30.62; sham: 72.56 ± 19.01), muscle relaxation was significantly increased after verum (85.82 ± 13.53), but not after sham SS-WBV (76.96 ± 4.85). Effect sizes using Cohen's d (80) on muscle relaxation in the verum condition was $d = -0.501$, and in the sham condition $d = -0.192$.

TABLE 2 Results of *t*-tests for each exercising condition.

	Verum SS-WBV 8.5 Hz ($n = 28$)				Sham SS-WBV 6 Hz ($n = 27$)				Intergroup Analysis for *E*	
	BL	*E*			BL	*E*				
	Mean ± SD	Mean ± SD	*t*	*p*	Mean ± SD	Mean ± SD	*t*	*p*	*t*	*p*
Musculoskeletal Well-being	75.64 ± 31.84	88.50 ± 11.54	−2.26	.032	80.41 ± 18.20	76.81 ± 26.15	.93	.359	2.13	.040
Muscle Relaxation	73.39 ± 30.62	85.82 ± 13.53	−2.21	.036	72.56 ± 19.01	76.96 ± 4.85	−1.16	.258	1.63	.109
Balance-Test (dominant leg)	74.81 ± 10.39	82.28 ± 11.76	−5.52	<.001	72.41 ± 7.28	76.90 ± 6.70	−6.01	<.001	2.08	.043
Balance-Test (non-dominant leg)	73.45 ± 7.03	80.05 ± 10.47	−4.52	<.001	72.82 ± 5.98	77.79 ± 7.57	−5.36	<.001	.92	.364
Sense of Balance	47.50 ± 6.12	54.71 ± 10.31	−3.82	.001	48.96 ± 6.50	56.78 ± 10.28	−4.69	<.001	−.74	.461
Surefootedness	48.86 ± 4.20	53.61 ± 9.96	−2.66	.013	51.22 ± 7.07	52.89 ± 8.43	−.80	.429	.29	.774
Flexibility-Test	3.77 ± 9.69	6.21 ± 8.61	−5.63	<.001	6.25 ± 11.03	7.82 ± 11.03	−6.51	<.001	−.60	.548
Sense of Flexibility	44.04 ± 8.47	57.11 ± 12.22	−4.55	<.001	48.48 ± 11.85	56.74 ± 13.40	−3.89	.001	.11	.916
Cognitive Interference (RT control)	145.96 ± 136.77	86.97 ± 101.34	2.14	.042	49.24 ± 439.90	137.36 ± 162.07	−.98	.335	−1.38	.171

Left: Verum sinusoidal side-alternating whole-body vibration (SS-WBV, 8.5 Hz) at Baseline (BL) and exercising condition (*E*); Middle: Sham WBV (6 Hz) at BL and E; Right: Intergroup Analysis for the exercising condition (verum SS-WBV *E* vs. sham SS-WBV *E*); Variables: Musculoskeletal well-being and muscle relaxation were assessed on a 100-point-rating-scale from 0 "not at all comfortable" to 100 "as comfortable as you can imagine"; Balance was measured with the modified Star Excursion Balance Test (mSEBT); Sense of Balance and Surefootedness were answered from 0 "a lot worse than usual" to 100 "much better than usual" and with 50 being "same as always"; Flexibility was assessed with the modified Fingertip to Floor Test (mFTF); Sense of Flexibility was answered from 0 "not at all flexible" to 100 "as flexible as I can imagine"; Cognitive Interference was calculated as the difference between reaction time (in milliseconds; ms) in incongruent trials minus the reaction time (ms) in control trials. Higher interference stands for lower inhibitory control; *p*-values are two-tailed with an α-level set at 5%.

Better balance after verum SS-WBV (H2)

A significant effect in balance (dominant leg) was found after verum SS-WBV ($t = -5.52$, $p < .001$, $N = 28$) and also after sham SS-WBV ($t = -6.01$, $p < .001$, $N = 27$). Compared to baseline measurement (verum: 74.81 ± 10.39; sham: 72.41 ± 7.28), the balance (dominant leg) increased significantly in verum (82.28 ± 11.76) and in sham SS-WBV (76.90 ± 6.70). Effect size using Cohen's *d* (75) on balance (dominant leg) in the verum condition was $d = -0.663$, and in the sham condition $d = -0.621$.

A significant effect in balance (non-dominant leg) was found after verum SS-WBV ($t = -4.52$, $p < .001$, $N = 28$) and also after sham SS-WBV ($t = -5.36$, $p < .001$, $N = 27$). Compared to baseline measurement (verum: 73.45 ± 7.03; sham: 72.82 ± 5.98), balance (non-dominant leg) increased significantly in verum (80.05 ± 10.47) and in sham SS-WBV (77.79 ± 7.57). Effect size using Cohen's d (80) on balance (non-dominant leg) in the verum condition was $d = -0.689$, and in the sham condition $d = -0.695$.

A significant effect in sense of balance was found after verum SS-WBV ($t = -3.82$, $p = .001$, $N = 28$), and also after sham SS-WBV ($t = -4.69$, $p < .001$, $N = 27$). Compared to baseline measurement (verum: 47.50 ± 6.12; sham: 48.96 ± 6.50), sense of balance increased significantly in verum (54.71 ± 10.31) and in sham SS-WBV (56.78 ± 10.82). Effect size using Cohen's *d* (80) on sense of balance in the verum condition was $d = -0.824$, and in the sham condition $d = -0.86$.

A significant effect in surefootedness was found after verum SS-WBV, ($t = -2.66$, $P = .013$, $n = 28$), but not after sham SS-WBV ($t = -0.80$, $p = .429$, $N = 27$). Compared to baseline measurement (verum: 48.86 ± 4.20; sham: 51.22 ± 7.07), surefootedness increased significantly in verum (53.61 ± 9.96), but not in sham SS-WBV (52.89 ± 8.43). Effect size using Cohen's d (80) on surefootedness in the verum condition was $d = -0.581$, and in the sham condition $d = -0.213$.

Better flexibility after verum SS-WBV (H3)

A significant effect in flexibility was found after verum SS-WBV ($t = -5.63$, $p < .001$, $N = 28$) and also after sham SS-WBV ($t = -6.51$, $p < .001$, $n = 27$). Compared to baseline measurement (verum: 3.77 ± 9.69; sham: 6.25 ± 11.03), flexibility increased significantly in verum (6.21 ± 8.61) and in sham SS-WBV (7.82 ± 11.03). Effect size using Cohen's *d* (80) on flexibility in the verum condition was $d = -0.261$, and in the sham condition $d = -0.137$.

A significant effect in sense of flexibility was found after verum SS-WBV ($t = -4.55$, $p < .001$, $N = 28$) and also after sham SS-WBV ($t = -3.89$, $p = .001$, $N = 27$). Compared to baseline measurement (verum: 44.04 ± 8.47; sham: 48.48 ± 11.85), flexibility increased significantly in verum (57.11 ± 12.22) and in sham SS-WBV (56.74 ± 13.40). Effect size using Cohen's *d* (80) on sense of flexibility in the verum condition was $d = -1.244$, and in the sham condition $d = -0.650$.

Less cognitive interference after verum SS-WBV (H4)

A significant smaller interference effect was found after verum SS-WBV ($t = 2.14$, $p = .042$, $N = 28$), but not after sham SS-WBV ($t = -0.98$, $p = .335$, $N = 28$). Compared to baseline measurement (verum: 145.96 ± 136.77; sham: 49.24 ± 439.90), the difference in reaction time between incongruent and control items decreased significantly after verum (86.97 ± 101.34), but not after sham SS-

WBV (137.36 ± 162.07). Effect sizes using Cohen's d (80) on cognitive interference in the verum condition was $d = 0.486$, and in the sham condition $d = -0.265$.

Discussion

Overall, promising effects were found for the verum WBV condition, but not for the sham condition, indicating acute musculoskeletal and cognitive effects of SS-WBV. More precisely, musculoskeletal well-being and muscle relaxation increased after SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz (H1). Also, flexibility assessed through the modified fingertip-to-floor method (mFTF) as well as through a single-item question increased after both conditions, 8.5 Hz as well as 6 Hz. (H2). Sense of balance, which was assessed with a single-item question only increased after SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz. However, balance measured with the modified Star Excursion Balance Test (mSEBT) improved after both SS-WBV conditions, 8.5 Hz and 6 Hz (H3). Lastly, cognitive interference measured with the Stroop Color Word Test decreased after SS-WBV with a vibration frequency of 8.5 Hz but not of 6 Hz (H4).

The aim of the current study was to conduct the acute effects of WBV training using a light and portable platform, incorporating the use of several physiological tests and questionnaires. Body balance performance measure included not only musculoskeletal outcomes, but also cognition. After one SS-WBV exercise, different physiological and cognitive measurements have shown improvement, with effect sizes for WBV training being small to moderate. Results might indicate that different variables could be sensitive for different vibration frequencies.

Whole-body vibration training is beyond physiological effects

WBV training has proven its health promoting effects in various outcomes such as higher musculoskeletal well-being, better flexibility and increased balance (17, 26, 27, 82, 83). Musculoskeletal well-being and flexibility improved after one session of WBV-training only in the verum group. This supports previous findings where it was shown, that WBV increases flexibility (63, 83), because vibration increases blood circulation and generates more heat, which facilitates flexibility. Additionally, WBV causes muscles to contract and relax, which may raise the pain threshold and could lead to participants being able to stretch further while experiencing less pain (84, 85).

Piecha et al. (2014) have shown that WBV increases postural stability (55), which allows us to move safely, which could be related to improved surefootedness. One reason for increased postural stability could be enhanced muscle strength (86). Thus, changes in muscle strength might play a significant role in increasing postural stability and should be addressed in future studies. An increase in surefootedness was significant in the verum group, but not in the sham group, indicating that participants walked more safely after higher vibration stimulation. Balance-Tests however showed not only the verum (8.5 Hz), but also the sham group (6 Hz) increased in balance. One could assume that this result might be related to a training effect on the balance test. However, this might also indicate that WBV also effects balance when light frequencies are applied, maybe because proprioceptive training does not need as high frequencies as e.g., relaxation and musculoskeletal well-being.

Having a good balance is associated with less falls (87), because sensorimotor performance is better (88). But this is not the only explanation. Less falls are also related to better cognitive performance (89, 90), especially with executive functions (91, 92), such as inhibition. For example, Hausdorff et al. (2005) measured inhibition with the Stroop Test in non-demented older adults and have shown that a lower performance in the Stroop Test was also linked to a lower gait performance (93).

Recent studies have shown that cognition may be enhanced through WBV training in mice and in humans (94). In their randomized controlled trial, Boerema and colleagues postulated that daily vibration trainings over 5-weeks improve motor performance and reduce arousal-induced home cage activity in mice (94). In humans, WBV training improved brain function tested with the Stroop Color-Word test. Accordingly, a recently published review from a Brazilian research group on the effects of WBV on different cognitive variables described cognitive enhancement through training and suggests more clinical trials to establish beneficial training parameters (95).

Findings of cognitive effects after WBV training are still rare and the underlying processes are not fully understood yet. Studies with mice have shown increased cholinergic activity after WBV (94). Also, cholinergic activity in humans is positively associated with Stroop Test results (96). Therefore, improvement of inhibitory control in humans may be due to enhanced cholinergic activity increased by WBV training. On the contrary, improved inhibitory control after (repeated) WBV training may be due to the connection of sensory brain regions and the prefrontal cortex. Sensory stimulation, as perceived while executing WBV training, enhances neurotransmission not only in sensory brain regions, but also in the prefrontal cortex (97). However, this finding might be unique for WBV compared to other forms of physical activity (e.g., walking), because Sanders and colleagues have not found any cognitive effects in older persons with dementia after participation in a walking and lower limb strength training program over 12 weeks (98).

Finally, the finding that inhibitory control may be improved through WBV could be an important implication for occupational stress research. Stress at work impairs inhibitory control (99), while inhibitory control is a personal resource that helps to deal with high work demands. Inhibitory control has shown to be related with mindfulness in early adolescence (100) and mindfulness has been shown to be a personal resource that reduces work stress in line with the job demands-resources model (101). According to Lee and Chao (2012), inhibitory control is important for psychological well-being and for achieving mindfulness, and therefore may help to reduce interference from emotional distractors (e.g., an angry face, a negative thought, or a negative event) (102). Thus, people may intentionally avoid emotional distractors and can focus on desired or goal-related information promoting their own well-being (103). This could be noteworthy for future studies exploring personal resources to cope with work demands, but also to reduce cognitions that are related with weaker musculoskeletal function, such as fear- avoidance beliefs, maladaptive back beliefs, and concerns of falling.

The cognitive enhancement was measured with the Stroop Color Word Test. Inhibition and therefore cognition improved from pre to post intervention. Interestingly, these effects were shown in healthy young participants, mostly students who are expected to already have a high level of attention. As in previous WBV-studies on inhibitory control (104–106), a Stroop Test was implemented immediately after the exercise. Future studies may also take long-term effects of SS-WBV on cognition into account.

Further studies may also focus on the aging workforce who could profit the most from SS-WBV interventions focusing on gait performance and frequency of falls, since these have been shown to be related with Stroop Test results (93). In line with this, training parameters concerning different outcome variables must be defined, so users understand which methods (e.g., SR-WBV, SS-WBV), frequencies, and training durations should be applied, if they not only want to increase bone density or reduce musculoskeletal pain, but also improve balance and inhibitory control.

Overall, SS-WBV has shown to be an appropriate way to improve different health-related outcomes. In this initial step, SS-WBV exercise has shown to increase inhibitory control in a young and healthy sample. Implemented as a worksite intervention, SS-WBV is expected to improve balance and reduce falls, especially in older workers.

Falls and cognitive interference

Research on falls and gait control differ between single falls and recurrent falls. On the one hand, single falls are known as accidental falls, and mostly due to extrinsic reasons, e.g., environmental or housing conditions (88, 89). On the other hand, recurrent falls are often usually based on intrinsic reasons, e.g., advanced age, diseases, or gait disorders (107). Recurrent fallers could profit from interventions such as preventive and therapeutic exercises, in order to improve mobility (108). Because only few effective treatment possibilities exist to effectively improve gait control and balance for fall prevention, new intervention possibilities must be explored (109).

SS-WBV is easily applied and has shown to be effective in improving balance among different studies (26, 63, 64). Interestingly, SS-WBV might affect balance in different ways: Firstly, SS-WBV might improve balance by a proprioceptive training of muscles (110). Secondly, SS-WBV might improve balance through relaxation of stiff muscles, and hence less weakened proprioceptive information in sensory tissues (111, 112), and pain inhibition (113). Thirdly, studies have shown increased inhibitory control, i.e., less cognitive interference, after WBV exercises. This might also indicate a contribution to the prevention of slip trip and fall incidents, because not only gait performance (93), but also falls (90–92) seem to be connected with Stroop Test results. However, underlying mechanisms need to be further explored to fully understand the relationship between WBV, inhibitory control, balance and falls.

Practical implications

In the current study, WBV exercises were applied on a transportable and manageable vibration platform which does not take up much space and time, because participants are not likely to sweat and would not need to change clothes or shower after WBV-training. Furthermore, WBV-training is very short. In the current study, three minutes of SS-WBV stimulation already showed positive effects. Due to these benefits and the positive physical and cognitive outcomes that were found in the current study, a next step could be to study SS-WBV health-interventions at work. Faes et al. (2018) found promising effects in increased balance and musculoskeletal well-being in office workers who spend a substantial amount of time in sitting position (26). Because several physiological and cognitive measures have been improved, positive effects are not only related to less falls, but may also be linked to better life satisfaction and personal well-being of employees, which could lead to more satisfied employees and better work performances (114, 115).

Limitations

The described effects of the current study were only observed directly after one training session. Further studies should address long-term effects of repeated SS-WBV trainings. Another limitation of the current study pertains to the “chosen” vibration frequencies. Because the lowest possible frequency (6 Hz) of the SS-WBV platform did not differ enough with the verum condition (8.5 Hz), effects on balance were observed in both conditions. To study effects on balance, the sham-group should possibly experience lower or no vibration frequencies, e.g., control group. However, no vibration frequency would carry the problem of the blindness of participants, because one might guess their group allocation when nothing happens. Finally, our study relates to a relatively young age of participants. Since slip, trip and falls are especially common in the elderly, further studies with older people are necessary.

Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Ethikkomission der Universität Bern. The participants provided their written informed consent to participate in this study.

Author contributions

FY designed the study, organized the experiment, analyzed the data, and wrote the manuscript, RSC helped design the study and wrote the manuscript, HML performed the measurements, analyzed the data, and wrote the manuscript, EA supervised the design of the study and analysis and wrote the manuscript. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Publisher's note

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"*Fine synergies*" describe motor adaptation in people with drop foot in a way that supplements traditional "*coarse synergies*"

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Synergy analysis via dimensionality reduction is a standard approach in biomechanics to capture the dominant features of limb kinematics or muscle activation signals, which can be called "coarse synergies." Here we demonstrate that the less dominant features of these signals, which are often explicitly disregarded or considered noise, can nevertheless exhibit "fine synergies" that reveal subtle, yet functionally important, adaptations. To find the coarse synergies, we applied non-negative matrix factorization (NMF) to unilateral EMG data from eight muscles of the involved leg in ten people with drop-foot (DF), and of the right leg of 16 unimpaired (control) participants. We then extracted the fine synergies for each group by removing the coarse synergies (i.e., first two factors explaining ≥85% of variance) from the data and applying Principal Component Analysis (PCA) to those residuals. Surprisingly, the time histories and structure of the coarse EMG synergies showed few differences between DF and controls—even though the kinematics of drop-foot gait is evidently different from unimpaired gait. In contrast, the structure of the fine EMG synergies (as per their PCA loadings) showed significant differences between groups. In particular, loadings for *Tibialis Anterior*, *Peroneus Longus*, *Gastrocnemius Lateralis*, *Biceps* and *Rectus Femoris*, *Vastus Medialis* and *Lateralis* muscles differed between groups ($p < 0.05$). We conclude that the multiple differences found in the structure of the fine synergies extracted from EMG in people with drop-foot vs. unimpaired controls—not visible in the coarse synergies—likely reflect differences in their motor strategies. Coarse synergies, in contrast, seem to mostly reflect the gross features of EMG in bipedal gait that must be met by all participants—and thus show few differences between groups. However, drawing insights into the clinical origin of these differences requires well-controlled clinical trials. We propose that fine synergies should not be disregarded in biomechanical analysis, as they may be more informative of the disruption and adaptation of muscle coordination strategies in participants due to drop-foot, age and/or other gait impairments.

KEYWORDS
electromyography, muscle synergies, non-negative matrix factorization, drop foot, gait

1. Introduction

Applying dimensionality reduction techniques to kinematic or electromyographic (EMG) data is a form of unsupervised learning (1, 2) to capture the lower-dimensional structure of the neural control of movement (1, 3–8). Independently on whether or not these "synergies" are of neural origin (4, 7), they are "descriptive" (8, 9) (in a mathematical sense) of the basis functions that best explain a high percentage of the variance in the data.[1] The investigator must first determine *a priori* if linear or nonlinear basis functions are most appropriate, and what is the discrete number of basis functions (i.e., synergies) that explain a "high enough" percentage of the variance (1). In practice, methods that produce linear basis functions are most popular such as Non-Negative Matrix Factorization (NMF) (5, 10), Principal Component Analysis (PCA) (11), Independent Component Analysis (ICA) (12), and Factor Analysis (FA) (13).

In the fields of biomechanics and neuromechanics, the number of synergies that together explain 80- 90% of the variance are considered sufficient to explain the dominant characteristics of the data and, therefore, most informative (3–8, 14, 15). We call these "*coarse synergies*." The residuals from the coarse synergies (i.e., which represent the remaining 20–10% of the variance) are, by construction, data (i) in which the investigator is *a priori* not interested (because they explicitly set the cut-off for variance explained), (ii) which cannot be accounted for by the linear model (a by-product of the preferred method (1)), or (iii) are considered noise (an assumption which must be proven) (16, 17). In either case, they are considered irrelevant or unimportant.

Here, we question this traditional interpretation of coarse synergies and the assumptions about their residuals to explore the subtle ways in which synergies can differ across populations. Our rationale is that there are coarse mechanical features of, in this case, locomotion that must be common to all participants—and are therefore not very informative of differences across populations. Therefore, we look to residuals as a more informative source of subtle differences.

In particular, here we focus on analysing the residuals after removing coarse synergies to establish whether or not they are irrelevant, and if they are informative of fine features of muscle coordination that are not captured by the coarse synergies. To do so, we apply dimensionality reduction to the residuals of the coarse synergies to extract "*fine synergies*." As a first example of this approach, we use EMG from leg muscles during locomotion to compare coarse and fine synergies between people with drop foot (DF) vs. unimpaired control participants (C).

2. Materials and methods

2.1. Participants

Two groups of people participated in this study. Ten individuals with clinically diagnosed unilateral drop foot without comorbidities that prevented locomotion formed the experimental group (DF). Their mean age was 52.9 ± 17.9 years, height 174.8 ± 9.1 cm, and body mass 68.8 ± 18.7 kg. The following medical diagnosis were represented: peroneal nerve palsy secondary to lumbar disc herniation ($n = 2$); post motor vehicle injury ($n = 1$); progressive muscular dystrophy ($n = 3$); surgical removal of a tumor at the level of the head of the fibula ($n = 2$); ischemic disease of the lower limbs surgically fitted with stents ($n = 1$); and, amyotrophic lateral sclerosis ($n = 1$). In daily life, all participants were ambulatory and did not report dependence on a wheelchair. During test day, they verbally declared a good health and physical condition to participate in the study. Sixteen unimpaired participants with a mean age of 25.3 ± 7.1 years, height of 176.6 ± 6.8 cm and body mass of 74.1 ± 10.5 kg constituted the control group (C). All participants gave their informed written consent to participate in this study. The procedures were approved by the Ethical Committee of the Medical Center of Postgraduate Education in Warsaw, Poland (84/PB/2016).

2.2. Instrumentation and data collection

Unilateral surface EMG (sEMG) was collected from eight muscles using a Noraxon system (Noraxon USA. Inc., USA). Data were collected from the involved limb of persons from the DF group, and from the right limb from control participants. The activity was recorded from the following eight muscles: *Tensor Fasciae Latae* (TFL), *Biceps Femoris* (BF), *Peroneus Longus* (PL), *Gastrocnemius Lateralis* (GL), *Vastus Lateralis* (VL), *Tibialis Anterior* (TA), *Vastus Medialis* (VM) and *Rectus Femoris* (RF). For each participant, the bipolar Ag–AgCl EMG electrodes (10-mm diameter, 20-mm dipole distance) location was identified according to guidelines for electrode placement developed by the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project and verified based on clinical muscle tests.

All participants walked barefoot and naturally at their self-selected speed along a 10m walkway. Trials with incidents were discarded from further analysis and the procedure was repeated. Two force plates (Kistler Holding AG, Switzerland) were used to determine ground reaction forces using Nexus 1.7.1 software, which afterwards was confirmed manually for each participant. Data was then exported to the Vicon Polygon system, which independently divided the gait into individual cycles and calculated the gait spatio-temporal parameters. EMG and Force plate systems were synchronized and had a sampling frequency of 1000 Hz. After data collection from the Drop foot group, kinetic and kinematic data were visually inspected to determine the results' homogeneity (**Supplementary Figure 6**).

2.3. Data analysis and muscle synergy extraction

Surface EMG signals were high-pass filtered to remove movement artifacts, using a third-order Butterworth high-pass filter at 20 Hz. On-line sEMG signals were displayed for inspection of the signal quality during measurement. The sEMG signals were rectified and smoothed with a 2 Hz second-order Butterworth low-pass filter to obtain the muscle contraction linear envelope. The third gait cycle from each participant was selected for analysis based on ground reactions forces data. The sEMG envelopes were processed into a time normalized sEMG profile (i.e., from 0 to

[1]That is, the original data can be approximated as a combination of the basis functions extracted from the original data.

100% of gait cycle, starting at heel strike). Next, each muscle's sEMG time series for each participant was normalized by the maximal peak value demonstrated by that specific muscle across gait cycles. Therefore, the magnitude of muscle activity was not taken into consideration in this temporal analysis.

Extraction of coarse synergies: We used the NMF algorithm to extract muscle synergies and their corresponding activation coefficients (i.e., weights) (10). This method calculates a set of synergy weights ($W_{m\times n}$) and synergy activations ($A_{x\times j}$), such that sEMG $= W \times A +$ residuals, where n is the number of synergies, m is the number of muscles (eight in this study), and j is equal to the number of sEMG data points (15). The residuals are defined as the difference between the experimental sEMG envelopes and the sEMG envelopes reconstructed from the product of the synergy weights and activations. The procedure to select the number of coarse synergies was to include as many as necessary to have $\geq$80% of variance accounted for (VAF) (15). To compare the coarse features of muscle coordination between control (C) and drop foot (DF) groups, we applied Statistical Parametric Mapping (SPM) to the reconstructed activity profiles, and a mixed design robust ANOVA with trimmed means (18) to compare the muscle weights extracted from the two coarse synergies that accounted for $\geq$80% of variance. The *spm1d* package (www.spm1d.org) was used to perform SPM analysis (19). SPM was used to compare the reconstructed muscles activity profiles between groups C and DF to detect whether the coarse synergies showed statistically significant differences over the gait cycle.

Extraction of fine synergies: To extract the residual sEMG signals, the above reconstructed signals were subtracted from the original experimental sEMG envelopes. PCA was applied to the residual components of EMG to extract the fine synergies for each participant in both groups. In contrast to the experimental sEMG envelopes that have a 0 floor and 1 ceiling—which NMF can accommodate best—the residuals are zero-mean time-series for which PCA is appropriate. For each participant, we extracted the principal components (PC's) and their loadings, which were then normalized based on the highest loading per participant for both groups (20).

To compare the fine features of muscle coordination between control (C) and drop foot (DF) groups, we also applied Statistical Parametric Mapping (SPM) to the reconstructed activity profiles, and a mixed design robust ANOVA with trimmed means to compare the normalized muscle loadings extracted from the fine synergies. Non parametric post-hoc analyses were used to compare individual muscle pairs when the results from the robust ANOVA revealed a main or interaction effect. All statistical procedures were performed with RStudio (RStudio Team, MA, USA).

3. Results

3.1. Spatio-temporal parameters

The spatiotemporal parameters of both groups are listed in Table 1, and were compared using t-tests for independent samples. Cadence for the DF group was 81.1 ± 2.42 steps/min, while the Control group was 90.6 ± 4.45 steps/min. Step length was 0.5 ± 0.07 m for the DF group and 0.66 ± 0.09 m for the Control group. Step width was 0.11 ± 0.02 for both groups. Stride time was 1.43 ± 0.14 (s) for DF and 1.29 ± 0.07 (s) for the Control group. Finally, walking speed was 0.8 ± 0.03 (m/s) for the DF group and 1.33 ± 0.06 (m/s) for the Control group. All spatiotemporal parameters were significantly different between groups ($p < 0.01$), except for Step Width (Table 1).

TABLE 1 Spatiotemporal gait patterns mean (± standard deviation) in drop foot (DF) and Control (C) groups.

	DF group	C group	p-Value
Cadence (steps/min)	81.11 ± 2.42	90.65 ± 4.45	0.0001
Step length (m)	0.5 ± 0.07	0.66 ± 0.09	0.0002
Step width (m)	0.11 ± 0.02	0.11 ± 0.02	—
Stride time (s)	1.43 ± 0.14	1.29 ± 0.07	0.0090
Walking speed (m/s)	0.8 ± 0.03	1.33 ± 0.06	0.0001

3.1.1. Coarse synergies

As expected, only two NMF factors sufficed to explain the gross features of muscle coordination in both groups (Supplementary Tables S2, S3). In the control group two factors explained an average of 88.1 ± 3% of variance accounted for (Supplementary Table S2 and Figure 1). Whereas for the drop foot group, the first two factors explained, on average, 91.52 ± 3.96% of variance accounted for (Supplementary Table S3 and Figure 1). We defined these first two factors that explain ≥85% to be the *coarse synergies* for the Control and Drop Foot groups.

The time histories of the coarse EMG synergies in the DF group showed few differences compared to Controls. While there are visual differences between the DF and Control groups, the only statistically significant ones (as per SPM, $p < 0.01$) occurred in the first *coarse*

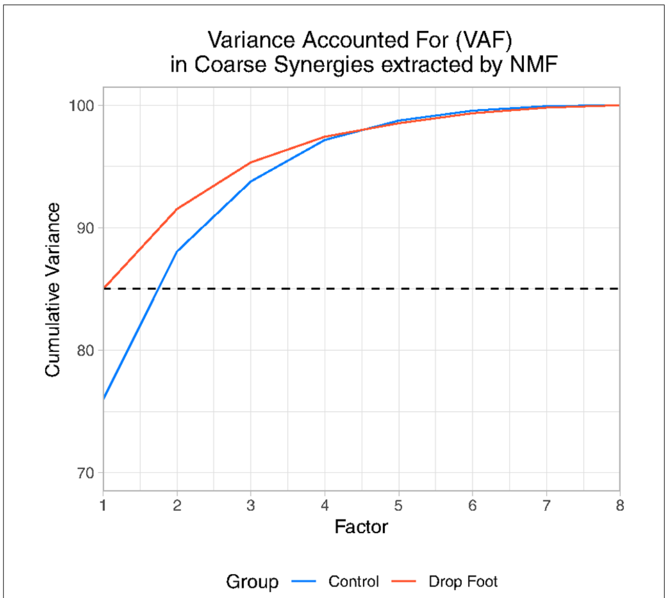

FIGURE 1
Cumulative variance accounted for each Factor extracted by NMF in drop foot (DF) and Control (C).

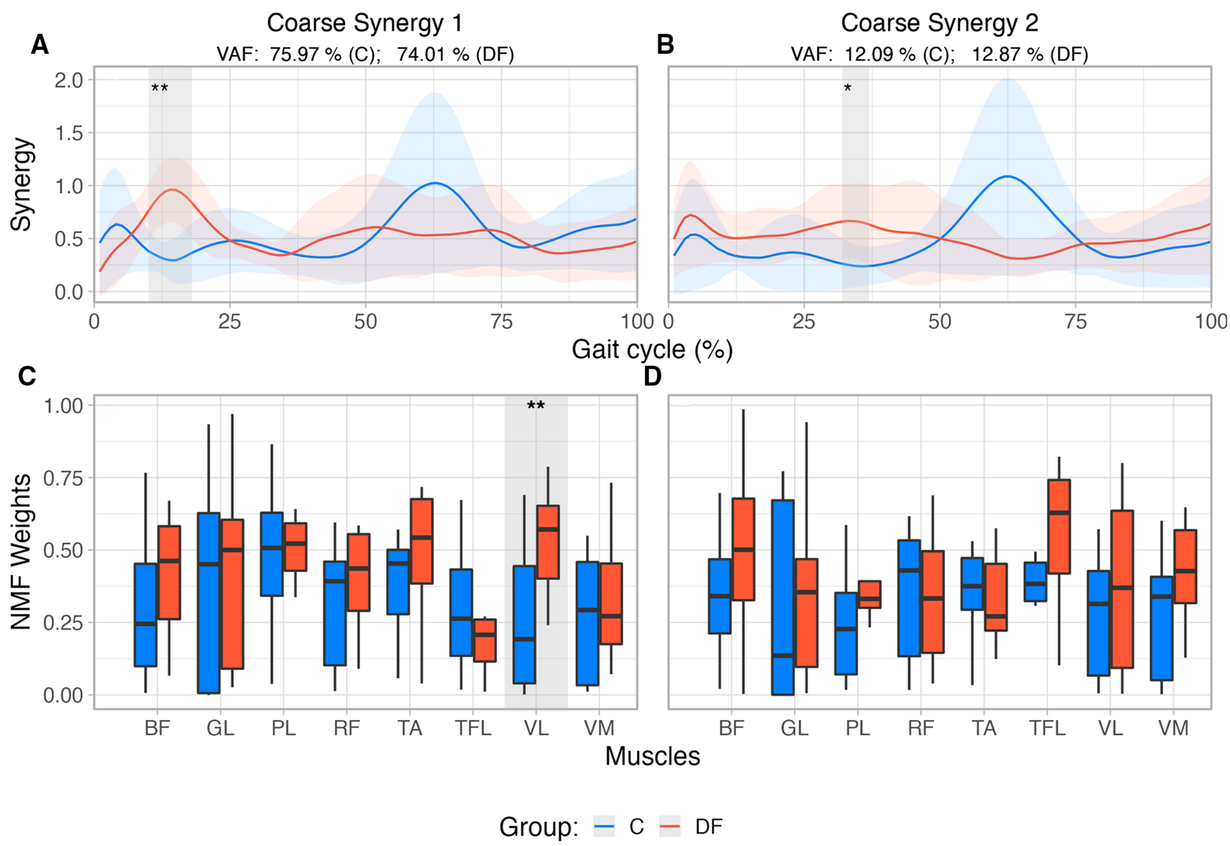

FIGURE 2
(A,B) Reconstructed muscle activity profiles based on weights extracted from first two *coarse synergies* for each group, accounting for >86% of variance in each group. Shaded areas identify differences between groups based on SPM{t} results and their corresponding levels of significance. (C,D) Coarse synergies muscle weights extracted from NMF for unimpaired control participants (C) and persons with drop foot (DF). *Significant at 5%; **Significant at 1%.

synergy from 10 to 18% of the gait cycle (**Figure 2A**). For the second *coarse synergy*, significant differences ($p = 0.015$) were only observed from 32% to 37% of the gait cycle (**Figure 2B**).

The structure of the coarse EMG synergies showed differences only for one muscle between the DF and Control groups. Muscle weights[2] extracted from NMF (**Figures 2C,D**) were compared using a Robust mixed effects ANOVA model. In the *first coarse* synergy, the analysis revealed a main effect for Muscle ($p < 0.01$), and Group ($p = 0.032$), with no interaction (Muscle × Group, $p = 0.3$). Post-hoc analysis revealed significant differences between groups for muscle VL ($p < 0.01$) only. Comparison of muscles weights extracted from the second coarse synergy did not show main effects for Muscle ($p = 0.07$), Group ($p = 0.05$) nor interaction (Muscle × Group, $p = 0.53$).

3.1.2. Fine synergies

Three *fine synergies* sufficed to explain ≥85% of variance in the residuals in both groups: 90.47% (±3.79 SD) and 90.46% (±3.24 SD) in the Control and DF groups, respectively (**Figure 3**).

SPM analysis did not reveal differences between groups at any level of significance in the histories of the three fine synergies (**Figures 4A,C**).

The structure of the first two fine synergies showed multiple statistically significant differences between the Control and DF groups, as per their loadings. Muscle loadings extracted from PCA (**Figures 4D,F**) were also compared using a Robust mixed effects ANOVA model, which revealed a main Group effect for the first and second *fine synergies* ($p < 0.01$, for both synergies), and a Muscle main effect ($p < 0.01$) in the second "fine synergy." Post-hoc analysis revealed statistical differences between both groups for muscles TA ($p = 0.016$), BF ($p = 0.038$), RF ($p = 0.049$), GL ($p = 0.015$), VL ($p = 0.01$) and VM ($p = 0.01$) in the first synergy, and PL ($p = 0.024$), RF ($p = 0.036$), TA ($p = 0.031$), and VM ($p = 0.036$), in the second *fine synergy*.

The third fine synergy did not show differences in its structure between Control and DF groups. The third synergy did not have a main Muscle ($p = 0.40$), Group ($p = 0.49$), nor interaction effect (Muscle × Group, $p = 0.52$). Moreover, all of their loadings tended to include or hover near zero. These results suggest the third fine synergy is likely unimportant to both groups (**Figure 4F**).

4. Discussion

Descriptive synergies which explain the majority of the variance in data (i.e., *coarse synergies*) are a common metric to compare

[2]In NMF factors are described by their "weights," whereas in PCA the term "loadings" is used.

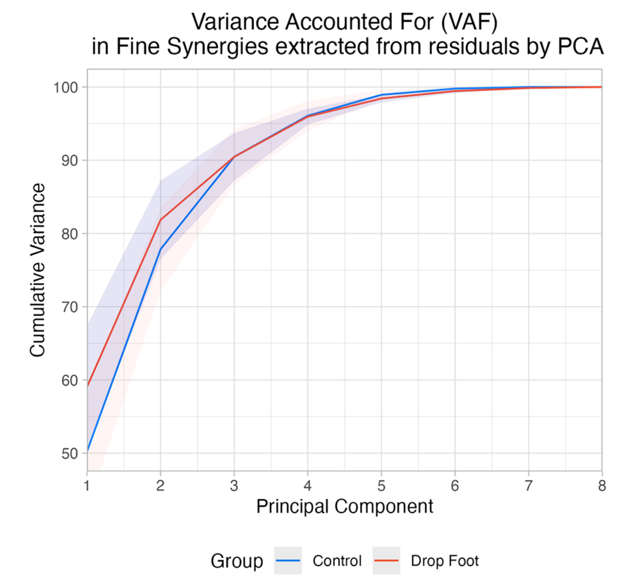

FIGURE 3
Cumulative variance accounted for each PC extracted from residuals by PCA in drop foot (DF) and control (C).

performance across populations. We argue that coarse synergies, in the case of DF at least, can be uninformative about differences between groups as they mostly capture the dominant biomechanical features of locomotion common to all participants. We thus explored the notion that descriptive *fine synergies* extracted from the residuals to the *coarse synergies* may be—by virtue of containing subtler features—more informative of differences across populations.

Our results show this is the case when analyzing EMG signals from control and DF participants as the fine synergies showed the most differences across populations—potentially revealing subtle disruptions and adaptations of muscle coordination strategies in participants with DF.

An important methodological aspect of our approach is that we first used NMF on the EMG data, and then PCA on their residuals. Our rationale is twofold. NMF is a well-founded approach for analyzing rectified and normalized EMG signals that lie between values of 0 and 1 due to the non-negative input constraint to perform factorization. As such, it is better suited to extract coarse synergies ($\geq$85% VAF) from processed EMG signals (10, 11, 21). The residuals of the EMG signals after removal of the coarse synergies are zero-mean by construction, and therefore PCA is the more appropriate technique for extracting fine synergies (1). We then focused on analyzing these residual EMG signals first and foremost to establish whether or not they had enough structure in their correlations to make them informative of fine features of muscle coordination that are not captured by the coarse synergies.

The nature of PCA loadings should be clarified before proceeding. PCA is a dimensionality reduction technique that

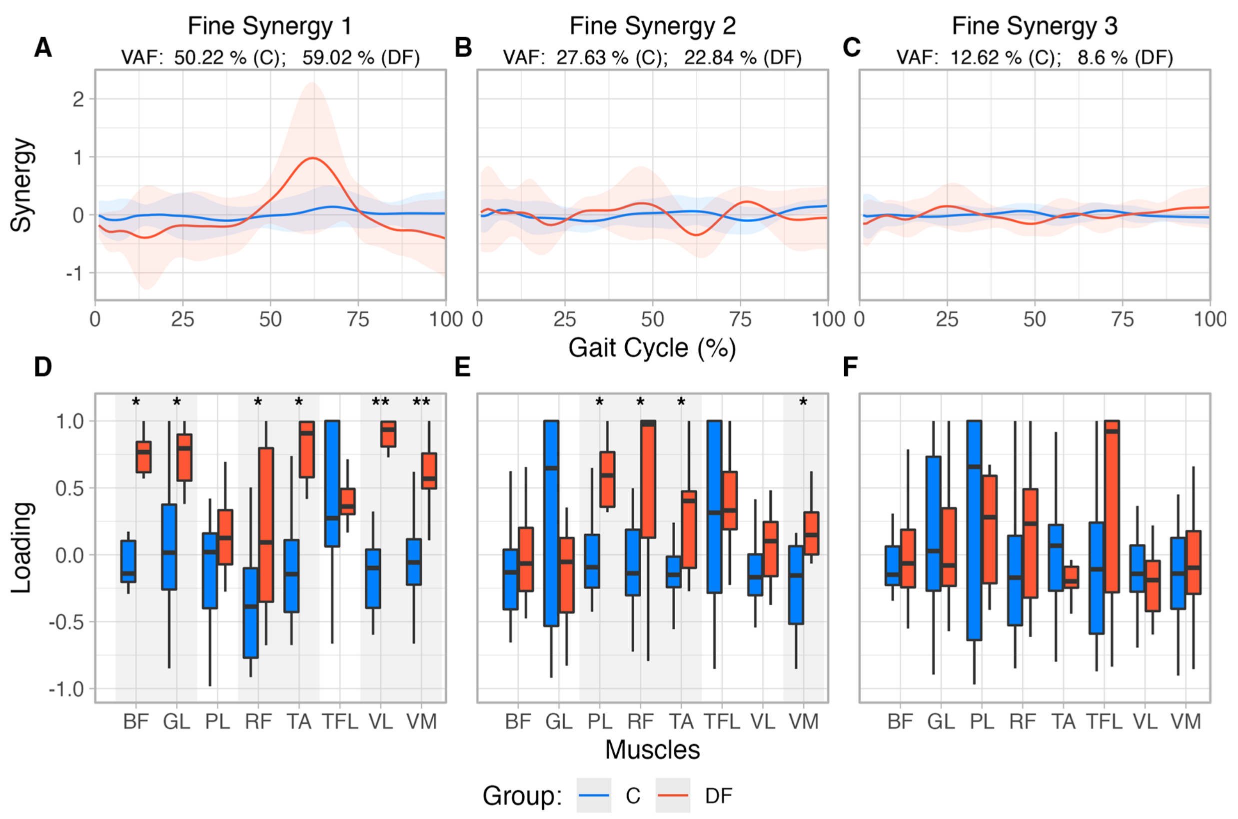

FIGURE 4
(A–C) Reconstructed muscle activity profiles based on loadings extracted from first three "fine synergies" for each group. (D–F) Fine synergies loadings extracted from PCA for unimpaired control participants and persons with drop foot. Also note the loadings in DF are in general closer to +1 in the DF case, indicating greater synergistic correlation among muscle activations.

approximates a high-dimensional signal with fewer basis vectors (PCs) that capture important features of correlations in the original signal. The values of PCA loadings have a range between −1 and 1, therefore, describe whether and how the elements of the original signal are correlated. Namely, loadings describe if there is structure to their correlations, or if their correlations hover near zero and therefore render the synergies informative. Importantly, PCA is obtained from the covariance matrix of the individual EMG signals, thus the correlations among EMG signals are what determine their loadings and not their overall level of activation. Therefore, a weakened muscle with a low level of activation—such as the TA in the DF group—can still have a loading close to 1 (or −1) in a PC if its activity is highly correlated (or anti-correlated) with the other muscles. On the other hand, a muscle could have a loading hovering near zero even if it is highly activated but uncorrelate with other muscles in that PC.

Given this preface, our results showed that the first two fine synergies in the DF participants were different from those in the controls. This is evidenced by the DF loadings being statistically different from controls in **Figures 4**, **5**. This is also valid for TA—even though we know it is weaker in the DF group—because its loadings are statistically different in the first and second fine synergies compared to controls. In contrast, all three fine synergies of the control participants, and the third fine synergy of the DF group, show little correlation structure as they loadings are hovering near zero. Therefore, those fine synergies are uninformative.

Dimensionality reduction techniques to extract coarse synergies have known limitations (1, 4, 9, 16). For example, synergies are necessarily descriptive of the correlations among muscle activities; but do not necessarily speak to the actual neural control producing the task (9). In addition, PCA relies on signal normalization, which for EMG is performed via maximum voluntary contraction (MVC) of each muscle. However, this process is unreliable and true maximal force is difficult to attain in individuals with motor deficits (22). Here, we normalized EMG signals based on the maximal activity of each muscle during the gait cycle. We did this to prove that weakness based on changes in the EMG signal is not the only change in muscle activity between groups, and is actually a change in the correlation structure among muscles that produces differences between groups. Since we already know that the activity levels will be different across groups, by normalizing to the maximal activity during the gait cycle we make the amplitudes of the signals comparable to reveal differences in the correlations among muscle activations. If scaling down the signals due to weakness is the only change during DF, we should not have found differences in the muscle loadings compared to controls. The presence of these differences in the fine synergies and not in the coarse synergies highlights the ability of fine synergies to reveal compensatory motor coordination strategies.

In order to test the usefulness of coarse vs. fine synergies to detect differences across groups, we compared the DF group to the so-called clinically neurotypical group. We consider young self-declared unimpaired people as such. On the other hand, if we had considered an age-matched group to those with DF, we would have the concern that they might exhibit some comorbidities of aging that would confound our results. Initially, 15 older subjects were

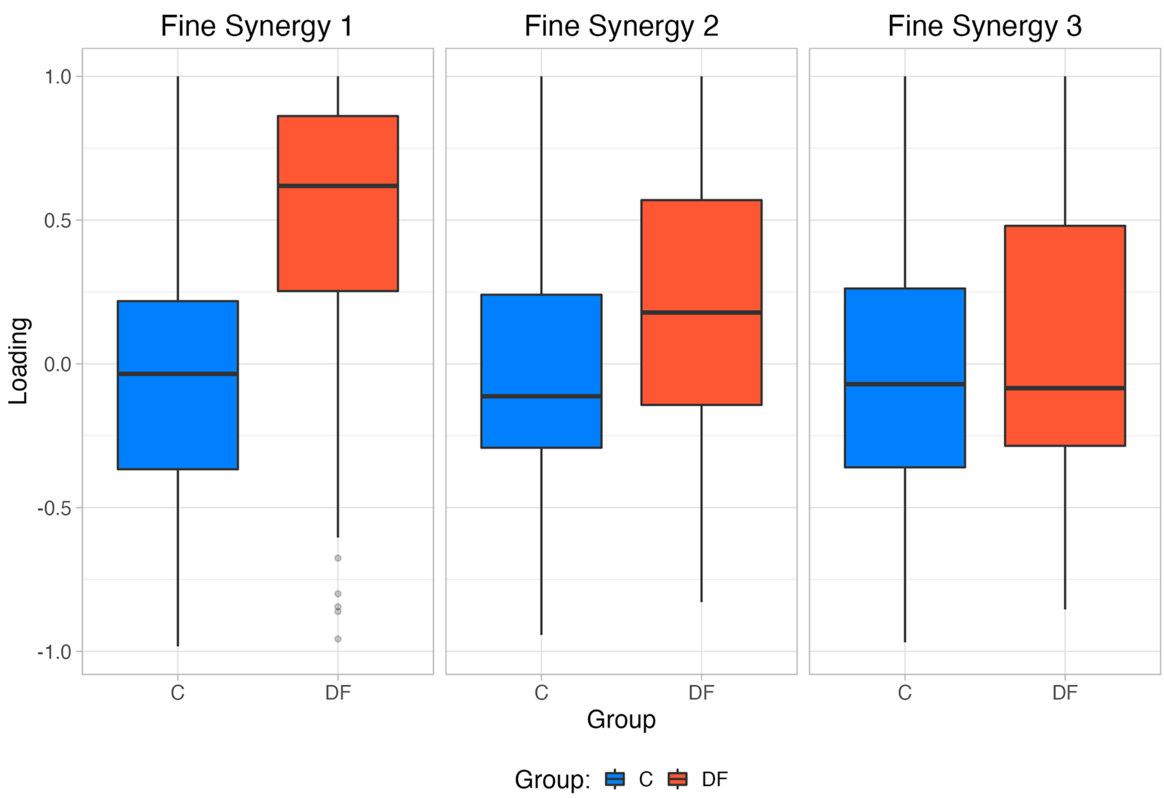

FIGURE 5
Mean Fine synergy loadings for each group extracted by PCA.

screened for enrollment in our study; however, they did not meet our inclusion criteria due to comorbidities. Thus we kept younger individuals as controls to avoid potential confounds of aging.

We recognize that our study had a small sample size, compared populations of different ages, and the number of electrodes may not fully capture the muscle activation patterns of the leg. However, to the best of our knowledge, aging does not affect kinetic and kinematic parameters during gait (23). While age could partly explain our results (or an interaction between age and DF), this could only be confirmed using a larger sample and a more complex experimental design that is beyond the scope of this work. Importantly, our goal was not to definitively declare DF from its various diagnoses, levels of impairment, clinical evolution (and/or age) as the main cause of differences between groups. We also do not claim that synergies of any kind can provide clinical insights unless and until they are used in the context of well-controlled clinical trials (which for DF is beyond the scope of this work). Rather, we used data from DF populations as a first example that allows us to question the traditional approach to, and interpretation of, descriptive 'coarse synergies' as biomarkers for changes in motor strategies. Our results show that changes due to DF (and/or aging) are not reflected in coarse synergies, further supporting the importance of analyzing "fine synergies"—which is the main topic and goal of our study.

To mitigate the limitations of our small sample size, we used robust inferential methods for hypothesis testing, which perform well with small sample sizes and when the assumptions of parametric statistics regarding normality and homoscedasticity are not met, and provide more accurate statistical results compared to classic parametric statistical techniques based on means comparisons (18).

From a technical perspective, our wired equipment limited the number of channels to record EMG signals from each participant to eight. We therefore chose to record the signals only from the affected side of each DF participant. Also, due to cable length, participants were only able to walk 10 m, the reason for which we analyze only the third cycle once they reached a stable gait pattern before starting to decelerate and come to a full stop. Therefore, we could not record EMG during three full strides at a participant's comfortable speed to assess recording's reliability (24).

Notwithstanding these limitations, we find that coarse synergies are not as informative of differences across populations during gait, as compared to fine synergies. In particular, we saw an increase in the correlation of the weakened TA muscle activation with other muscles in the DF group (i.e., higher loading value), which was also seen in most of the recorded muscles in the first and second fine synergies, with only *Tensor Fasciae Latae* not being statistically different in any synergy (**Figure 4**). In the DF group, the increased loading for the *Biceps Femoris* may act as a compensatory mechanism to decrease hip flexion during initial contact, potentially translating to a decreased step length. Additionally, the increased loading for the *Vastus Medialis* and *Lateralis* could represent a mechanism to decrease knee flexion during midstance. These changes have been previously reported in people with DF during ground clearance and foot-ground interaction (25). Previous findings have also shown that the presence of weakness during foot dorsiflexion in DF activates compensation strategies and influences muscle force and activation distribution (26). It was found that reduced forces of individual muscle groups of the ankle joint are compensated for by the increased strength of others acting on this joint (i.e. *Tibialis Posterior*, *Gastrocnemius Lateralis*), along with other muscles in neighboring joints (i.e. *Biceps Femoris*, *Rectus Femoris*, *Vastus Lateralis*, *Tensor Fasciae Latae*) (26). Considering that we found differences in PCA loadings within the same muscles (with the exception of *Tensor Fascia Latae*), our results from the fine synergies could reflect the same gait adaptations in the DF group as previously described.

Our results have allowed us to better characterize motor deficits and adaptations in persons with DF, based on differences in fine synergies as compared to control participants. This highlights the importance of considering not only the dominant features of a behavior (coarse synergies), but also the fine details revealed by fine synergies.

Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Ethical Committee of the Medical Center of Postgraduate Education in Warsaw, Poland (84/PB/2016). The patients/participants provided their written informed consent to participate in this study.

Author contributions

MB and FVC contributed to conception and design of the study. AB, MB, MA and RN pre-processed the data, AB performed the statistical analysis. AB, MB, HA, RN and FVC interpreted the results. AB and FVC wrote the first draft of the manuscript. AB, MB, HA and RN wrote sections of the manuscript and created the figures. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2023.1080170/full#supplementary-material.

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Prioritizing limb loading improves symmetry during dual-tasking in individuals following anterior cruciate ligament reconstruction

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Understanding the extent to which attention prioritization interfere with limb loading in daily activities following anterior cruciate ligament reconstruction (ACLr) is important for reshaping loading behaviors. A dual-task paradigm, prioritizing limb loading symmetry (LLS) during standing or response time during an upper extremity task response time task was used to probe the effects of attention prioritization of loading. Individuals 115.6 ± 17.8 days post-ACLr (ACLr; $n = 13$) and matched healthy individuals ($n = 13$; CTRL) performed a simple response time (RT) task and 2 dual tasks prioritizing limb loading (LS-RT) and response time (RT-LS). 2 × 3 General Linear Model repeated measures analyses determined effects of group and focus condition on LLS error and response time. Significant interaction ($P = 0.010$) was noted in LLS error. ACLr group, exhibited greater LLS error in RT ($P = 0.001$) and RT-LS ($P = 0.001$) than LS-RT condition. ACLr group exhibited greater LLS error in the RT ($P = 0.001$) and RT-LS ($P = 0.040$) than CTRL, but not in LS-RT. A main effect of condition ($P < 0.001$) for response time indicated that times were slower in LS-RT compared to RT ($P < 0.001$) and to RT-LS ($P < 0.001$) for both groups. These data suggest that limb loading symmetry during standing is more automatic for controls than individuals following ACLr. Unlike controls, improving loading symmetry during standing requires additional attention in individuals in early recovery following ACLr.

KEYWORDS
ACLR, dual-task, limb loading symmetry, automaticity, rehabilitation

Introduction

Biomechanical studies have shown that individuals following ACLr adopt loading strategies that shift mechanical demand away from the surgical knee and limb during functional and athletic tasks (1–6). These strategies are most apparent during bilateral tasks that require equal distribution of weight across both limbs. A recent study found that at 3 months post-ACLr, individuals underloaded their surgical limb during standing, sit-to-stand and squat tasks by as much as 24% when they were not specifically attending to task performance (7). Despite specific emphasis on increasing loading of the surgical limb during postoperative rehabilitation, these deficits appear to persist over time. Longitudinal assessments of loading indicate that asymmetrical loading observed during squatting at 3 months post-ACLr does not improve at 5 months (8). It is suggested that traditional rehabilitation is not sufficient for the restoration of limb and joint loading. When considered along with studies that report similar deficits during squatting in individuals up to 22 months post-surgery (3) and during landing 2–3.5 years post-surgery

(6, 9), it is clear that these early underloading strategies persist long-term. The persistence of a generalized asymmetrical loading strategy is of concern as a prospective assessment of athletes following ACLr found that the odds of suffering a second ACL injury were 2.3 times greater in those who exhibited asymmetrical knee loading during a drop land at the time they returned to sports (10). Asymmetrical ground reaction forces during landing have been prospectively linked to risk for ACL injury in healthy individuals (9). In addition, asymmetrical loading has been attributed to the progression of knee osteoarthritis (11, 12).

There is evidence to suggest that asymmetrical loading strategies observed in early rehabilitation are not the consequence of an inability to accommodate loading demands, but a strategy carried over from early adaptations to joint level impairments experienced following injury and surgery. Individuals 3 months post-ACLr are able to improve loading symmetry by up to 14% during standing, sit-to-stand and squatting tasks when they were instructed to focus on distributing loads evenly through the limbs (7). These improvements from natural loading to instructed loading conditions suggest that increasing loading of the surgical limb during functional tasks may require additional attention at this time post-surgery.

Evaluations of individuals 1-year post-ACLr that show greater cortical activation during motor accuracy tasks and increased postural errors in dual-task conditions, support this premise. When compared to healthy controls, greater cortical activation (electroencephalogram data) was observed during a quadriceps force reproduction task in individuals 12.0 ± 4.7 months post-ACLr in order to achieve the same accuracy (13). The authors suggest that individuals following ACLr require more focused attention to accomplish the same motor task involving the knee joint than healthy individuals. Moreover, the introduction of a cognitive task during a single-limb balance task increased balance errors in individuals 14 months post-ACLr compared to controls (14–17).

Currently, it is not known if maintaining limb loading symmetry during common daily activities requires additional attentional resources for individuals following ACLr. If individuals require more cognitive resource to achieve appropriate loading in early rehabilitation, one might expect that the effects of exercises performed in rehabilitation may not carry over to daily activities with different attentional prioritizations (18, 19). Understanding the extent to which attention prioritization interferes with loading in a common daily activity in early rehabilitation is important for reshaping early loading behaviors.

Therefore, the purpose of this study was to use a dual-task experimental paradigm to determine the effects of attention prioritization of maintaining limb loading symmetry; comparing the performance of an upper extremity response time task and the degree of loading symmetry under conditions with different attentional prioritizations. It is hypothesized that during the dual-task condition that involves two performance goals (loading symmetry and response time), improved limb loading symmetry and increased response time will be observed in the condition where attention is prioritized to loading symmetry but not in the condition where attention is prioritized to response time in individuals following ACLr; however, these tradeoffs between loading symmetry and response time will not be observed in healthy individuals.

Materials and methods

Two groups of participated in this study: individuals 115.6 (17.8) days post anterior cruciate ligament reconstruction (ACLr; $n = 13$) and healthy controls (CTRL; $n = 13$). Participants' descriptive information is reported in **Table 1**. The participants in the ACLr group were recruited from four physical therapy clinics in the greater Los Angeles area. They were enrolled in the study if they were (1) between the ages of 14–50, (2) 10–16 weeks status post ACLr, (3) currently participating in physical therapy, and (4) cleared to perform the experimental tasks. Participants in the control group were recruited to match the participants in the ACLr group based on age- (±2 years), sex-, height-, weight-, and physical activity (Spots Activity and Function form, Cincinnati Knee Rating System). Control participants were excluded if they reported: (1) prior or current ligamentous or meniscal injury or surgery on lower extremities, (2) current or history of pathology or morphology in lower extremities that could cause pain or discomfort during physical activity (contralateral limb; ACLr group), and (3) any pathology or medical condition that may impair their ability to perform the tasks proposed in this study.

Procedures

Testing took place at the University of Southern California, Division of Biokinesiology and Physical Therapy's Human Performance Laboratory located at the Competitive Athletes Training Zone (CATZ) in Pasadena, CA. All procedures were explained to each participant and informed consent was obtained as approved by the Institutional Review Board of the University of Southern California, Health Sciences Campus. Parental consent and youth assent were obtained for participants under the age of 18 years. After consenting, participants completed the

TABLE 1 Participants characteristics.

	ACLr ($n = 13$)	CTRL ($n = 13$)
Age (years)	24.6 (9.8)	24.3 (9.2)
Sex	5 M/8 F	5 M/8 F
Height (cm)	1.71 (0.08)	1.71 (0.08)
Weight (kg)	71.66 (9.25)	70.8 (9.05)
Days post-ACLr	115.6 (17.8)	
Graft type (n)		
Bone-Patellar Tendon-Bone autograph	8	-
Hamstring autograft	1	-
Allograph	4	-
Physical Activity	95.00 (27.12)	96.15 (12.44)
IKDC overall	58.5 (11.69)	99.5 (1.30)

Values presented as mean (standard deviations) unless otherwise indicated.

subjective portion of the International Knee Document Committee (IKDC) form and Cincinnati Sports Activity and Function form to determine their current functional status and physical activity prior to injury, respectively. Age, height, weight, dominant limb (defined as leg the participant would kick a ball with), and knee medical history were recorded.

Task

Participants were asked to perform an upper extremity response time (UERT) task under three different attentional conditions. For this task, participants stood on two separate force platforms (BTS P-6000:BTS Bioengineering Corp, Milan, Italy) with their feet shoulders width apart in front of a 4' × 4'light board (Dynavision D2^{TM} Visuomotor Training device, Dynavision International LLC, West Chester, OH, USA; **Figure 1**). The light board made up of 64 targets arranged in 5 concentric rings. The board was positioned at a distance in front of the participant so that they were able to reach all the targets on the most peripheral ring without side-to-side trunk movement. The targets were divided in four quadrants consisting of 18 targets each. The top two quadrants were used for the UERT task (**Figure 1** upper left corner, solid and dashed rectangles). Participants were instructed to respond as fast as possible to depress a target when it illuminated. After the first illuminated target, each target depressed signaled the illumination of the next target with a latency of 0.02 s. Each UERT trial was performed for a total of 60 s.

Attentional conditions

The UERT tasks were performed under three attentional conditions.

Response time only (RT): The response time only condition was introduced first to probe individual's natural loading strategy when performing the UERT task. In this condition, no instructions were given regarding weight bearing and participants were not informed that they were standing on force platform or that ground reaction forces were being recorded. This condition required participants to focus only on response time of the UERT. Prior to testing participants were given the following instructions: "Tap the illuminated targets as fast as you can during the task." Performance feedback of the UERT task was provided to the participants after each trial. For the next two conditions individual were given two tasks and asked to perform both tasks but prioritize their attention to one of the tasks.

Prioritize limb loading symmetry (LS-RT): Participants were instructed to perform the UERT task while loading their limbs symmetrically. In the LS-RT condition, they were asked to prioritize maintaining limb loading symmetry (LS) while responding to the illuminated targets as fast as possible (RT). They were given the following instructions: "Distribute your weight evenly on the platforms and tap the illuminated targets as fast as you can. In this task, it will be more important that you distribute your weight evenly on the platforms as accurately as possible".

Prioritize response time (RT-LS): In the RT-LS condition, they were asked to prioritize responding to the illuminated targets as fast as possible while maintaining limb loading symmetry. They were given the following instructions: "Distribute your weight evenly on the platforms and tap the illuminated targets as fast as you can. In this task, it will be more important that you tap the lights as fast as you can."

After performing three trials in the RT condition, LS-RT was introduced followed by the first RT-LS condition. The second and the third LS-RT and RT-LS trials were then introduced alternatively (**Figure 2**). To avoid a learning effect, the illuminated targets were presented in a random order for each trial and participants were given 3 practice sessions to familiarize them with the light board. For each condition, 3 trials (60 s each) were used for analysis. Performance feedback of the UERT task was also provided to the participants after each trial in these two conditions.

Data analysis

Vertical ground reaction force (GRF) and response time were collected during each UERT task. Vertical GRF of each limb was measured using two separate force platforms. Response time, defined as the amount of time from when the light was illuminated to when the light was tapped by the participant, was output from the Dynavision D2^{TM} in seconds to two decimal places.

Vertical GRF impulse was calculated as the area under the vertical ground reaction force time curve during task execution using custom Matlab program (Mathworks, Natrick, MA, USA). Limb loading symmetry (LLS) during each UERT task was calculated as a between limb ratio of vertical ground reaction force impulses using Equation 1. To calculate the LLS in the CTRL group, limbs were matched to the ACLr group based on dominance regardless surgery.

$$\frac{\text{surgical limb (matched limb dominance in CTRL)}}{\text{non} - \text{surgical limb (matched limb dominance in CTRL)}} \quad (1)$$

LLS of 1 indicates equal distribution of weight between the limbs, LLS less than 1 indicates loading of the surgical/matched limb was less that the non-surgical/matched limb; and LLS greater than1 indicates loading of the surgical/matched limb is greater than the non-surgical/matched limb. To determine the degree of limb loading error in each condition, LLS error was calculated as the absolute value of 1-LLS (|1-LLS|) For all conditions, averages of LLS error and response time across 3 trials for each condition were used for analysis.

Statistical analysis

A priori sample size analyses on primary variables of interests (LLS error and response time) were performed using pilot data collected on 10 subjects (ACLr, $n = 5$ and control, $n = 5$). Data from the pilot study suggested that the results were normally distributed. Sample size calculations for group and prioritization condition comparisons using independent- and paired-samples

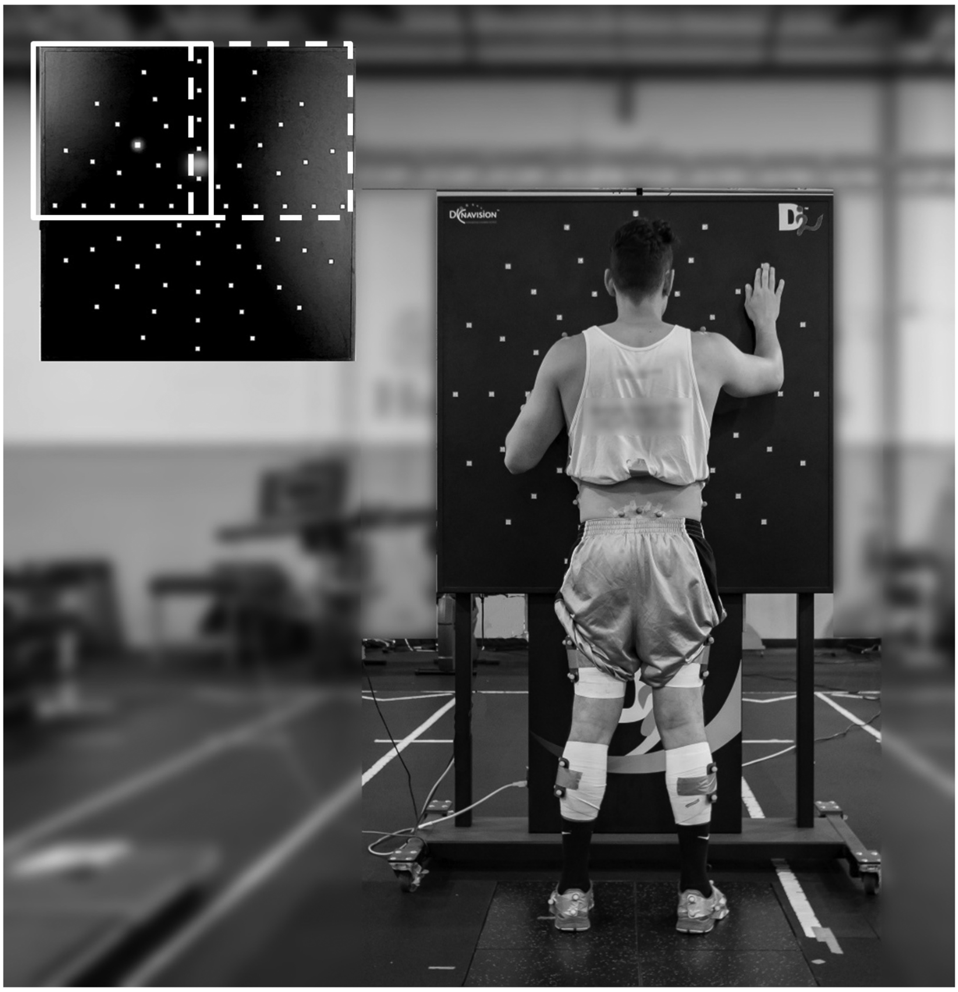

FIGURE 1
Experimental set-up for the UERT task and the dynavision D2 system. The dimension of the light board is shown at the upper left corner. The white solid square indicates the upper left quadrant and dashed square indicates the upper right quadrant used in the UERT task.

Practice (3 trials)	RT (3 trials)	LS-RT	RT-LS	LS-RT	RT-LS	LS-RT	RT-LS

FIGURE 2
Testing sequence.

t-tests indicated that a minimum of 4 participants per group were needed to detect the expected differences in limb loading symmetry between prioritization conditions (Cohen's $d = 1.85$ and power $= 0.81$) and response time (Cohen's $d = 2.38$ and power $= 0.87$) in the ACLr group with an alpha level of 0.05.

Separate 2 (Group) × 3 (Prioritization) General Linear Model (GLM) repeated measures analyses were performed to assess the effects of group and focus prioritization on limb loading symmetry and response time. In the case of a significant main effect or interaction, planned comparisons using independent- or paired-samples t test were conducted to compare limb loading symmetry and response time between groups and focus conditions. Significance level for all the tests was set at $\alpha = 0.05$ (IBM SPSS Statistics, Version 22, IBM Corp., Chicago, IL).

Results

LLS error

For LLS error, a significant interaction ($F = 5.68$, $P = 0.01$) between group and prioritization was noted (**Figure 3**).

When comparing within the ACLr group, LLS error was significantly greater in the RT only condition ($P = 0.001$; Table 2) and RT-LS condition ($P = 0.01$; Table 2) compared to the LS-RT condition. No significant difference was noted among conditions when comparing LLS error within the CTRL group (RT vs. RT-LS, $P = 0.76$; RT vs. LS-RT, $P = 0.95$; RT-LS vs. LS-RT, $P = 0.82$, Table 2).

When comparing between groups, the ACLr group exhibited significantly greater LLS error in the RT ($P = 0.001$, Table 2) and RT-LS ($P = 0.04$, Table 2) conditions compared to the CTRL group, but not in the LS-RT ($P = 0.985$, Table 2) condition.

Response time

For response time, a main effect of prioritization was observed ($F = 24.95$, $P < 0.001$, Figure 4). When collapsed across group, response time was significantly slower in the LS-RT condition compared to the RT only ($P < 0.001$; Table 3) and to the RT-LS ($P < 0.001$; Table 3) conditions. Response time in the RT-LS condition was not significantly different compared to the RT condition ($P = 0.469$; Table 3).

Discussion

Understanding the extent to which attention prioritization influences limb loading symmetry in individuals following ACLr is particularly important at this stage of recovery. This represents a time in which individuals are re-establishing their loading behaviors in rehabilitation and are increasing their daily activities. As highlighted by previous studies, despite their ability to perform more symmetrically with focused attention, individuals 3 months post-surgery utilize strategies that shift the load away from the surgical limb during tasks that mimic daily activities (7). Using a dual-task paradigm, the current study demonstrates that when compared to non-injured controls, attaining limb loading symmetry may require additional attention in individuals 3 months post-ACLr.

Insight into spontaneous or natural limb loading distribution during standing was provided in the RT condition, as participants were not aware that symmetrical limb loading was a goal or that it was being measured. With explicit focus on performing the UERT task as fast as possible, control participants exhibited relatively symmetrical limb loading with an average of 6% of LLS error. However, LLS error during the RT was 16% in the ACLr group, highlighting a natural tendency to underload their surgical limb. Performance of the UERT test was similar between groups.

The results of the dual-task comparisons suggest that concurrent tasks influence individuals following ACLr differently that non-injured controls. Central processing capacity is limited; during a dual-task condition this capacity is shared between two concurrent tasks (20). If a greater proportion of processing capacity is required by the prioritized task, there is less available capacity to allocate to the concurrent or secondary task. If the secondary task requires more capacity than it is available,

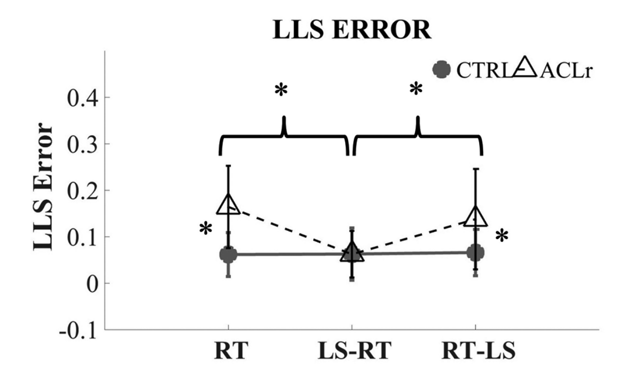

FIGURE 3
Limb loading symmetry (LSS) error across groups and focus prioritizations.

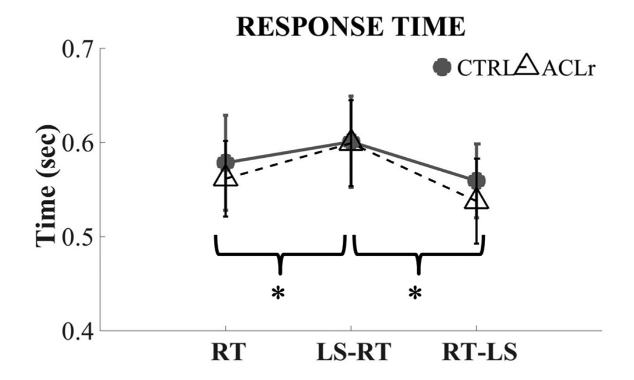

FIGURE 4
Response time groups and focus prioritizations.

TABLE 2 The effects of group and prioritization on LLS error, and pairwise comparisons across three focus conditions.

LLS error	ACLr			CTRL			*P*-value		
	RT	LS-RT	RT-LS	RT	LS-RT	RT-LS	Main effect (group)	Main effect (focus condition)	Interaction
Mean	0.164	0.062	0.138	0.061	0.062	0.066	0.014	0.010	0.010[a,b,c,d]
Standard deviation	0.088	0.050	0.108	0.047	0.056	0.049			

[a]ACLr RT vs. ACLr LS-RT.
[b]ACLr RT-LS vs. ACLr LS-RT.
[c]ACLr RT vs. CTRL RT.
[d]ACLr RT-LS vs. CTRL RT-LS.

TABLE 3 The effects of group and prioritization on response time, and pairwise comparisons across three focus conditions.

Response time (seconds)	ACLr			CTRL			*P*-value		
	RT	LS-RT	RT-LS	RT	LS-RT	RT-LS	Main effect (group)	Main effect (focus condition)	Interaction
Mean	0.56	0.60	0.54	0.58	0.60	0.56	0.469	<0.001[a,b]	0.211
Standard deviation	0.04	0.05	0.05	0.05	0.05	0.04			

[a]RT vs. LS-RT.
[b]RT-LS vs. LS-RT.

performance of that task will degrade. This is seen in the degradation or increase in response time in the dual-task condition (LS-RT) by both groups. In this condition loading symmetrically was prioritized in the instructions. Once participants were asked to focus on loading symmetry and prioritize this goal, performance of the secondary task degraded. Slower response times were observed in the LS-RT compared to the RT condition in both groups suggesting that focusing on loading symmetry depleted the central processing capacity and interfered with UERT performance. However, the fact that participants in both groups demonstrated similar increases response time suggests that the cognitive resources needed to attend to loading symmetry did not differ between groups.

When prioritizing LLS both groups exhibited symmetrical loading with only 6% of LLS error. This is particularly important in the ACLr group, as the single task condition suggests that their natural tendency is to underload their surgical limb. Limb loading error improved from 16% in during the RT to 6% in the LS-RT condition in the ACLr group. These data suggest that improved loading symmetry is achievable when performing a concurrent task if individuals specifically prioritize their loading behaviors.

When individuals prioritized performance of the UERT as fast as possible (RT-LS), response time was similar to the RT only condition in both groups. However, when asked to prioritize response time, loading symmetry degraded in individuals post-ACLr compared to when limb loading was prioritized. On average, 14% error in loading symmetry was observed in the ACLr group compared to 6% in the control group. While the LS-RT condition suggest that focusing on loading symmetry requires additional attention for both groups, the ability to maintain loading symmetry in the RT-LS condition in the controls suggests that loading symmetry may be, to certain extent, an automatic response for non-injured individuals. Automaticity is often conceptualized as the ability to perform a task with minimal cognitive demands and minimal interference from other concurrent information processing (18). The focus on RT did not influence loading symmetry in controls indicating that loading symmetry is a natural or automatic posture that requires minimal cognitive demands. The greater LLS error observed in the ACLr group indicates that achieving loading symmetry is not automatic and requires additional attention that was not being prioritized.

These findings have direct implications for the rehabilitation post-ACLr. Individuals in early recovery may need to prioritize loading symmetry during dual-task training until this posture becomes more automatic. Moreover, the inability to maintain loading symmetry when a concurrent attention is present and prioritized suggests that individuals following ACLr may more readily adopt this underloading strategy during daily activities. During daily activities, individuals often perform more than one task at a time. It is likely that the loading goal emphasized in the rehabilitation sessions may not be carried over as a priority into daily living. As such, the loading practice during daily activities may serve to reinforce the asymmetrical loading strategy which may underlie the persistence of the asymmetrical behavior. Therefore, the present study supports the inclusion of dual- or multi-taking training stimuli during rehabilitation especially during tasks that mimic daily activities.

Study limitations

Given that standing may be less challenging than other tasks performed throughout the day, the influence of distracted attention on limb loading behaviors may be underestimated in the is study. Interpretation of these data is limited to individuals 10–14 weeks post-ACLr. Self-reported IKDC scores were consistent with those reported in similar cohorts in other studies (18–20), indicating that individuals in the ACLr group were recovering typically. However, it is not known how these results would apply to those who have less typical recoveries. These data do not allow for speculation of how the demands of loading symmetrically change over time. Further work is needed to determine how training can reduce the cognitive demands of loading in the population.

Conclusion

The present study demonstrates that, in contrast to uninjured controls, maintaining limb loading symmetry during standing is not a natural or spontaneous loading strategy for individuals in early stages of recovery following ACLr. However, when loading symmetry was prioritized, individuals following ACLr are able to achieve typical symmetry demonstrated in healthy individuals. When a second task is introduced and prioritized, this improved symmetry is not maintained indicating that maintaining loading symmetry requires additional attention following ACLr. For controls, the ability to load symmetrically while attending to another task indicates that symmetrical limb loading may be a more automatic response that requires minimal cognitive resources. The present study supports the inclusion of dual- or multi-task training with a focus on loading symmetry during early recovery following ACLr.

Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Institutional Review Board of the University of Southern California, Health Sciences Campus. Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

Author contributions

M-SC contribution includes study design, data collection, data analysis, data interpretation, and drafting of the manuscript. SS contribution includes study design, data interpretation, and drafting of the manuscript. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of stochastic resonance whole-body vibration on sensorimotor function in elderly individuals—A systematic review

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Introduction: Due to demographic changes, falls are increasingly becoming a focus of health care. It is known that within six months after a fall, two thirds of fallers will fall again. Therefore, therapeutic procedures to improve balance that are simple and can be performed in a short time are needed. Stochastic resonance whole-body vibration (SR-WBV) may be such a procedure.
Method: An electronic search to assess the effectiveness of SR-WBV on balance in the elderly was conducted using databases that included CINAHL Cochrane, PEDro, and PubMed. Included studies were assessed using the Collaboration Risk of Bias Tool by two independent reviewers.
Results: Nine studies showing moderate methodological quality were included. Treatment parameters were heterogeneous. Vibration frequency ranged from 1 to 12 Hz. Six studies found statistically significant improvements of balance from baseline to post measurement after SR-WBV interventions. One article found clinical relevance of the improvement in total time of the "Expanded Time to Get Up and Go Test".
Discussion: Physiological adaptations after balance training are specific and may explain some of the observed heterogeneity. Two out of nine studies assessed reactive balance and both indicated statistically significant improvements after SR-WBV. Therefore, SR-WBV represents a reactive balance training.

KEYWORDS
stochastic resonance therapy, whole-body vibration (WBV), falls, aged, postural balance [MeSH]

Introduction

Ageing is associated with sensorimotor deficits resulting in muscular weakness, mobility issues, balance disorders and in gait disorders, and this leads to falls and reduced independence in everyday life (1). The sensorimotor system constantly and unconsciously regulates its movements and postural control based on perceived information to achieve postural stability. Numerous research studies have assessed the effect of stochastic resonance (SR) stimulation to the lower extremity on postural regulation and balance performance in sub-populations such as healthy adults (mean age: 23.04 years, ±6.33 years) (2), elderly individuals (mean age: 73.00 years) (3), or individuals with comprised health suffering from Parkinson (4), or multiple sclerosis (5) or stroke (6, 7), SR has been shown in a variety of physiological systems (8–11), in which the presence of noise below the sensory threshold could enhance the response of the system to weak signals (12–14).

Collins et al. (12) postulated that SR could be used to elderly individuals with elevated sensory thresholds.

Whole body vibration (WBV) with stochastic resonance (SR) can easily applied. SR-WBV does not lead to exhaustion and blood pressure and lactate levels are low during vibration training (15). SR-WBV could be easily personalized to the individual's level of fitness (16). For example, elderly with low baseline fitness who want to start an exercise program should start with a program that meets their physical capabilities (16). Older people with frailty or pre-frail condition are advised to undergo a "skilling-up" phase before undertaking more traditional forms of training (17). SR-WBV can be used as a training modality for the "skilling-up" phase (16, 18, 19). Compared to traditional balance training, there are indications on how to design a training regime (sets, rest between sets, session per weeks, etc.) and on the other hand the training protocols are characterized by a short duration between 1 and 5 min of intermittent or continuous WBV application (20). There is no need to change clothes or shoes or to shower afterwards, which might be important in the working world or for adults who do not want to waste time on intensive training (2). Eichelberger et al. (21) were able to determine a decrease in accelerations with increasing distance from the vibrating plate due to damping properties of the involved body structures. However, it is known that a prolonged exposure to vibration (e.g., driving, hammering) may lead to musculoskeletal and neurological disorders (22, 23). Systematic reviews and meta-analysis (24–27) have shown that shorter exposure to vibration have a positive effect on muscle strength and postural control if the training regime (e.g., amplitude, duration and frequency of vibration) is correctly dosed.

SR-WBV differs from sinusoidal WBV in that the stimuli are randomized and amplified using noise (25, 28). This results in a generation of action potentials by the suprathreshold stimuli (29). SR-WBV induces an excitatory stimulus to the alpha motoneuron *via* mono- and polysynaptic pathways and elicits muscle activation in response, resulting in body stabilization (30). SR-WBV can be understood as reactive balance training that simulates a fall situation itself through the application of unpredictable, random, and multidirectional displacements of the stance surface (31). Reactive balance training means that a person has the ability to react to a loss of balance, because reactive balance is a key factor that ultimately determines whether an individual will sustain a fall (32). Reactive balance can be profoundly impaired in older adult populations (32).

In contrast to SR-WBV, sinusoidal WBV are constant. If the stimulus remains the same, the body adapts very quickly and this slows down the impact of growth stimulus (27). Three WBV devices were used in clinical settings: sinusoidal vertical (SV-WBV), sinusoidal side-alternating (SS-WBV), and stochastic resonance (SR-WBV). While the sinusoidal WBV devices uses a single plate for standing, the SR-WBV device uses two plates for standing (24, 33–36). Due to the different physiological mechanisms of impact and use of equipment, this paper focuses on SR-WBV.

Furthermore, study results demonstrated that whole-body vibration training provides more than physiological effects (2, 37). Animals study showed that daily exposure to WBV over five weeks significantly improved cognition in young mice compared to non-vibrated mice (38, 39). Regterschot et al. (37) could determine that passive WBV could improve executive functions in healthy young adults. They postulated that WBV has the potential as a cognition-enhancing therapy. Chan et al. (40) reported that executive functions are a set of cognitive processes that regulate, manage and control other cognitive processes in order to achieve a goal, such as planning, mental flexibility, multi-tasking etc. Research findings described that cognitive decline and falls are linked (41–43) and that cognitive training improve balance and gait (44).

Aim

A systematic literature review on the effects of SR-WBV on postural control have been conducted previously (27). As the number of publications on SR-WBV has increased significantly in recent years, this present systematic literature review aims to provide an update on the status quo of the efficacy of SR-WBV on postural control in frail elderly individuals. The research question was: could SR-WBV positively influence postural control in individuals with balance disability?

Methods

Study design

This paper is an update of the systematic review by Rogan et al. (27). In advance, a registration on PROSPERO (CRD420203194) was conducted and the guideline "Preferred Reporting Items for Systematic Reviews and Meta-Analyses" (PRISMA) was used for reporting. This current systematic review used the same methodological approach as the first study. The inclusion and exclusion criteria were identical. The same search terms were used on the same databases. The data collection process was more comprehensive in this study. Besides the training load, the intervention protocol and the measurement instrument tools were now included. The risk of bias was assessed with the same instrument (The Cochrane Collaboration Cochrane Risk of Bias Tool) as in the first study.

Information sources

Electronic searches were conducted on CINAHL, Cochrane Central Register of Controlled Trials, Physiotherapy Evidence Database (PEDro), and PubMed up to August 2022. In addition, a hand search of the reference lists of included studies, research institution websites, and Google Scholar was conducted.

Search strategy

The PICO model was used in this study. The PICO acronym stands for **P**opulation (elderly, frail elderly), **I**ntervention (WBV exercise), **C**omparator (no treatment, or other balance exercise), Outcomes (postural control, static, dynamic, functional balance). Search terms included: (i) "stochastic resonance whole-body vibration" OR "SR-WBV" OR "stochastic vibration" OR "stochastic training" AND (ii) "balance" OR "postural control* " OR "postural stability".

Eligibility criteria

This study included intervention studies and randomized controlled pilot studies. German- and English-language articles with intervention and control groups from the fields of geriatrics were considered. For studies with frail elderly persons, those aged 65 years and older were eligible. Studies with frail elderly persons under 65 years of age, studies with elderly persons with "fit" status, and studies with neurological diseases were excluded.

Data collection process

Two independent study nurses screened and analyzed the title and abstract for inclusion and exclusion criteria. In the next step, the full text was read and included in this systematic literature review if eligible. For each included article, authors, population, intervention protocol, outcome parameters, results, and training load were extracted and electronically recorded by two independent study nurses.

Study risk of bias assessment

The Cochrane Collaboration Cochrane Risk of Bias Tool (RoB) (45) was used to assess the internal validity of the included articles. Two independent reviewers (SR, JT) assessed the methodological quality of the eligible studies with "The Cochrane Collaboration's tool for assessing risk of bias". The criteria list comprised six items and each item were scored with + for yes, with—for no, and with? if the information was not provided or was unclear. A study was determined as having a low risk of bias if all criteria are fulfilled with yes. A study has a moderate risk of bias when one or more items are scored with unclear, while a study has a high risk of bias if one or more key domains have been rated with no. The level of agreement between the independent reviewers who rated the primary studies was 98%.

Results

Study selection

There were 1,206 matches of studies. Of these, 262 duplicates were removed. A total of 944 titles and abstracts were screened, and 917 articles were removed due to systematic reviews articles ($n = 8$), application of sinusoidal vibration ($n = 898$), application of stochastic vibration *via* the sole of the foot or knee ($n = 7$), effects of SR-WBV on postural control or pain ($n = 2$), pelvic floor muscle ($n = 2$). The remaining 27 full texts were read, of which 9 articles were included in this systematic review (**Figure 1**). Three articles originated from Germany (4, 46, 47), and six from Switzerland[1] (16, 19, 30, 36, 48, 49). Six trials were designed as pilot study (16, 19, 30, 36, 48, 49) and three as randomized controlled trials (4, 46, 47).

Balance survey method

A total of six articles examined static balance (16, 19, 30, 47–49), six studies examined dynamic balance (4, 16, 19, 30, 47, 49) while five studies observed functional balance (19, 46–49).

Result overview of the studies

Overall, four of nine studies showed statistically significant balance improvements within the SR-WBV group in the before-after comparison (4, 19, 46, 48). **Table 1** summarizes the findings of the individual articles.

Training loads

In four of the nine studies, participants received a single training session with SR-WBV (4, 19, 30, 46). The remaining five studies determined the effect of SR-WBV after multiple interventions. The range was 12–36 training sessions (16, 36, 47–49).

In three studies (4, 46, 48), the frequency was increased and the starting position on the vibration device was progressively adjusted to the participants. All three studies use 5 sets and 60 s of vibration. One study did not specify a frequency (47). In three trials, continuous vibration was performed at a frequency of 5 Hz and five series with a duration of 60 s and rest of 60 s (19, 36). Four studies applied a frequency of 6 Hz, with 5–6 series of 60 s duration and 60 s rests (4, 16, 30, 46, 49).

The control group (CG) received no active intervention in four out of nine studies (19, 46, 47, 50). In one study, the CG completed a different intervention (4). Sham intervention was performed in four other studies. **Table 2** gives an overview of the training load.

The evaluation of the methodological quality was classified as followed (**Table 3**): one study did not use the method of allocation concealment (47), seven studies (4, 16, 19, 30, 46, 47, 49) did not report the blinded status of the investigator or participant, and four studies (4, 19, 46, 49) showed incomplete outcome data. They presented only change percentage data. They did not give any information about baseline and intervention data and no effect size calculation was used. **Table 3** provides an overview of the risk of bias of the included studies.

Discussion

During aging, sensory symptoms such as absent reflexes are clinically relevant. They are not only debilitating but also responsible for changes. This systematic literature review addressed the research question "could SR-WBV positively

[1]These articles are from the authors of this systematic review.

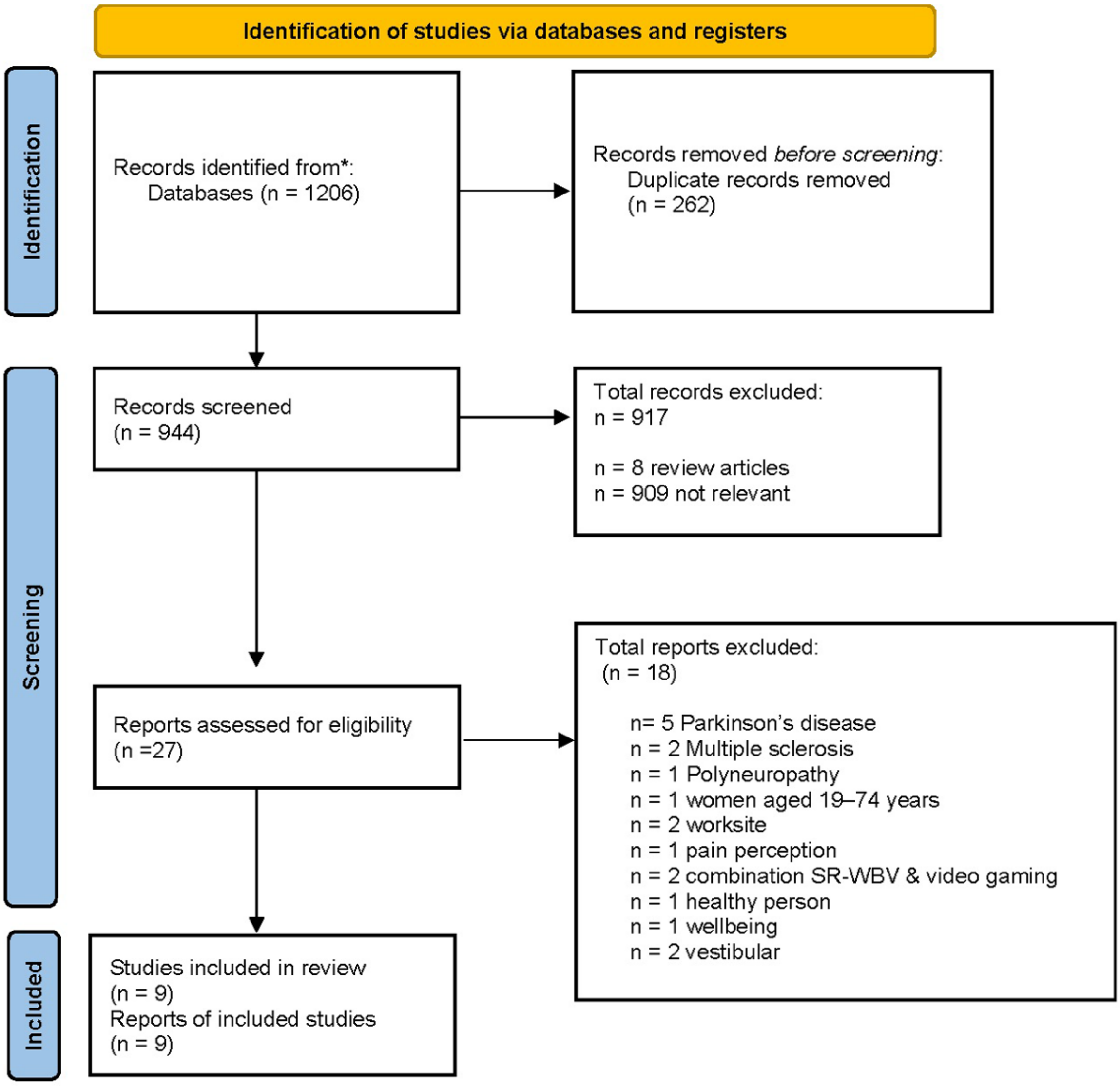

FIGURE 1
Flow chart of the search.

influence postural control in individuals with balance disability?" using published results.

In summary, the effect of SR-WBV on postural control presents a mixed picture. The statistically significant results from four studies were contrasted with five statistically unsound results. However, the effect size was strengthened by Rogan et al.'s (49) indication of clinical relevance. They were able to demonstrate clinical relevance (2.9 s) for the Expanded Timed Get-up-and-Go (ETGUG) test (51) in frail individuals after SR-WBV training. The SR-WBV group was a median of 3 s faster after the intervention period compared with the baseline measurement ($P = 0.043$; ES: 0.91). This study result has immediate consequences in terms of treatment recommendation for frail individuals with a postural control deficit of dynamic balance (52). It is known that a normal sensory system is necessary for successful postural control and movement. The central nervous system must accurately assess the position of the body in space and the limbs in relation to each other (proprioception). Postural reflexes must be released efficiently when external perturbations are detected. Maintaining balance must be automatic so that it is not impaired by other tasks. During aging, impairments of the sensorimotor system lead to a loss of postural control and to falls. The process of postural control depends on many sensory signals and neurological pathways and maintaining the quality of these systems at their optimal level is fundamental. SR-WBV could play an important role toward addressing postural control, by involving an interaction of different types of neurophysiological sensors and the adaptation of afferent and efferent signals, the SR-WBV likely serves as an exercise for the sensorimotor system. Tan et al. (53) demonstrated in their systematic review a significant positive benefit on postural control (SMD = 0.61, 95% CI: 0.12 to 1.09, $P = 0.01$) and muscle activity in tibialis anterior (SMD = 0.46, 95% CI: 0.04 to 0.88, $P = 0.03$) and gastrocnemius (SMD =

TABLE 1 Overview of the study characteristics of the included studies.

Study	Population	Intervention protocol	Target parameters		Results	
	Group: SR-WBV; intervention group (IG); control group (CG), (n), sex (men (m)/ women (w)), mean years (±SD)		Biomechanical measurement method	Functional parameter/outcome	Effect within group (effect size: ES)	Effect between groups
Dittrich et al. (47)	Prospective controlled study SR-WBV: $n = 41$ (m: 13/ w: 28) CG: $n = 52$ (m: 13 / w: 39) SR-WBV: m: 69.1(±4.0) w: 66.4(±5.0) CG: m: 68.1 (±6.8)/ w: 70.0 (±7.5)	WBV: Exercises on ZeptorMed (4 exercises per training session from a pool of 15; individual, not listed which exercises). IG: activity in everyday life was not changed	Static & dynamic balance with Biodex Stability System	Dynamic, functional balance with motoric assessment according Runge [Chair Rising, Timed-up-and-Go (TUG), Tandem walk, etc.]	IG (women) Chair Rising significantly decreased by 0.9 s ($p = 0.003$; ES = 0.4) TUG significantly reduced by 0.4 s ($p = 0.000$; ES = 0.4)	
Haas et al. (46)	RCT cross-over Group A: $n = 34$ Group B: $n = 34$ m: 53/w: 15 65.0 years (±7.8)	SR-WBV: Free two leg stand with shoes, with slightly bent knees CG: rest for the same duration		Static and functional balance with UPDRS motor scores und Sit-to-Stand (STS)	Significant reduction in UPDRS motor score ($p < 0.01$) after SR-WBV Group A: - 16.8% Group B: - 14.7%	
Kessler et al. (48)	RCT pilot study SR-WBV: $n = 13$ (m: 5/w: 8) CG: $n = 11$ (m: 3/w: 8) SR-WBV: 90.7 years (±7.5) KG: 83.8 years (±9.3)	SR-WBV: Parallel stance (increase possible without holding), tandem stance, slow dynamic squats CG: Sham therapy without increasing 1 Hz		Static and functional balance with chair rise Test during Short Physical Performance Battery test (SPPB)	Chair rising significantly reduced ($p = 0.001$; ES r = 0.89)	Significantly higher SPPB score significantly for SR-WBV compared to CG ($p = 0.035$; ES r = 0.43)
Rogan et al. (16)	RCT crossover pilot study Group A: $n = 10$ Group B: $n = 10$ Group A: 76.8 years (±7.7) Group B: 80.7 years (±5.7)	Parallel stance with shoes with slightly bent knees	Static balance with Kistler force plate	Dynamic balance with functional reach test FRT; Expanded Timed get Up and Go Test (ETGUG); (Single-task / Dual-task)		
Rogan et al. (49)	RCT crossover pilot study SR-WBV: $n = 9$ (m/w: n.i.) SR-GKV: 88.5 Jahre (±6)	Parallel stance with shoes with slightly bent knees		Static, dynamic and functional balance with semitandem/tandem stand & Chair Rise Test (during SPPB Test), ETGUG	Large ES in SPPB score after SR-WBV ($p = 0.039$)	Large ES for ETGUG ($p = 0.043$)
Rogan et al. (30)	RCT crossover pilot study SR-GKV: $n = 9$ (m:4/w:4) SR-GKV: 88.5 Jahre (±5.9)	Parallel stance with shoes with slightly bent knees		Static, dynamic balance by ETGUG, chair rising	Large ES for SPPB ($p = 0.121$); ETGUG ($p = 0.011$);	
Rogan et al. (19)	RCT crossover pilot study SR-WBV: $n = 10$ (m: 5/w: 5) CG: $n = 10$ (m:5/w:5) SR-WBV: 80.2 years (±6.8) CG: 77.4 years (±7.1)	SR-WBV: Parallel stance with shoes with slightly bent knees KG: Sham intervention with 1 Hz; same position as SR-WBV		Static, dynamic and functional balance with semitandem/tandem stand & Chair Rise Test (during SPPB Test), ETGUG		

(*Continued*)

TABLE 1 Continued

Study	Population	Intervention protocol	Target parameters		Results	
	Group: SR-WBV; intervention group (IG); control group (CG), (n), sex (men (m)/ women (w)), mean years (±SD)		Biomechanical measurement method	Functional parameter/outcome	Effect within group (effect size: ES)	Effect between groups
Rogan et al. (36)	RCT crossover pilot studie SR-WBV: $n = 10$ (m/w: n. i.) CG: $n = 10$ (m/w: n.i.) SR-WBV: 76.8 years (±7.7) KG: 80.7 years (±5.7)	SR-WBV: Parallel stance with shoes with slightly bent knees KG: Sham intervention with 1 Hz; same position as SR-WBV	Functional balance by chair rising on a Kistler force plate		Significantly faster rising ($p = 0.09$)	
Turbanski et al. (4)	Case-control-study SR-WBV: $n = 26$ (m/w: n.i.) CG: $n = 26$ (m/w: n. i.) $n = 52$ (m: 38/w: 14) $n = 69.1$ years (± 8.9)	SR-WBV: Upright standing on ZeptorMed CG: 15 min walk	Dynamic balance by means of TS and narrow PS on moving platform		Tandem stance was significantly longer after SR-WBV over 20 s ($p = 0.04$)	

0.68, 95% CI: 0.14 to 1.23, $P = 0.01$) using sinusoidal whole-body vibration in individuals with a sensorimotor deficit after ankle injury. They concluded that whole body vibration has the potential to improve sensorimotor deficits involving balance, strength, joint position sense, and muscle activity in people with chronic ankle instability. However, Lesinski et al. (54) formulated in their systematic review and meta-analysis article a balance training regime for healthy elderly by a training period of 11–12 weeks, a training frequency of three sessions per week, a total number of 36–40 training sessions, a duration of 31–45 min of a single training session, and a total duration of 91–120 min of balance training per week. Comparing these findings with findings from young healthy adults, it seems plausible that almost the same balance protocols are effective in healthy young and older adults and there seems to be no age effect (54). In this current article, no of the included articles reported this amount of training regimes. Fisher et al. were able to illustrate in their meta-analysis, that long-term WBV (between 4 weeks and 32 weeks) could significantly improve functional balance (Timed-up-Go test: SMD = −0.18; 95% CI: −0.32, −0.04; 10 min walking test SMD = −0.28; 95% CI: −0.56, −0.01). However, no significant changes were found in elderly individuals (tinetti gait scores: SMD = 0.04; 95% CI: −0.23, 0.31, 6 min walking test: SMD = 0.37; 95% CI: −0.03, 0.78).

TABLE 2 SR-WBV training load as used in the different studies under investigation.

Study	Duration / (sessions per week)	Frequency	Sets, duration, rest
Dittrich et al. (47)	12 weeks / (3×)	n. i.	3 × 45–60 s, 3 s
Haas et al. (46)	1 day / (1×)	6 Hz (±1 Hz)	5 × 6 s, 6 s
Kessler et al. (48)	4 weeks / (3×)	3–6 Hz	5 × 6 s, 6 s
Rogan et al. (16)	2 × 4 weeks / (3×)	5 Hz	5 × 6 s, 6 s
Rogan et al. (49)	2 × 4 weeks / (3×)	6 Hz	5 × 6 s, 6 s
Rogan et al. (30)	1 day / (1×)	6 Hz	6 × 6 s, 6 s
Rogan et al. (19)	1 day / (1×)	5 Hz	5 × 6 s, 6 s
Rogan et al. (36)	4 weeks / (3×)	5 Hz	5 × 6 s, 6 s
Turbanski et al. (4)	1 day / (1×)	6 Hz (±1 Hz)	5 × 6 s, n. i.

n. i., no indication; h, hertz; s, seconds.

It is known that muscle strength is a potentially important factor contributing to postural control (61). Large effects of strength training could be determined for static and dynamic balance in elderly individuals, but only a small effect was found for dynamic balance in young adults (62). Son et al. (63) were able to demonstrate that strength training increase muscle strength in ankle musculature and improve one-leg-standing balance compared to control situation. It can be concluded that the intensity of strength training is fundamental not only for increasing muscle strength but also for improving postural balance in elderly participants.

Furthermore, Kingwell described that exercise has the potential to improve cognitive function (64). Explanatory models address the fact that WBV stimulate mechanosensory receptors (e.g., tactile corpuscles). These signals are transmitted to the primary somatosensory cortex. These areas have connection with region in the prefrontal cortex that strongly involved in cognitive processing (37, 65). An indirect pathway involves the limbic system (e.g., amygdala and hippocampus, important areas of learning and memory), which can mediate the effects of sensory correlations on the prefrontal cortex (66). The amygdala has projections to non-thalamic nuclei (e.g., the cholinergic nuclei of the basal forebrain) with diffuse connections to several brain

TABLE 3 Risk of bias of the included studies.

	Random sequence generation (selection bias)	Allocation concealment (selection bias)	Blinding of participants (performance bias)	Blinding of personnel (detection bias)	Incomplete outcome data (attrition bias)	Selective reporting (reporting bias)	Other bias
Dittrich et al. (47)	?	–	–	+	?	?	–
Haas et al. (46)	?	?	–	+	–	+	–
Kessler et al. (48)	+	+	–	–	+	+	+
Rogan et al. (30)	+	+	+	?	?	–	+
Rogan et al. (16)	+	+	+	–	+	+	+
Rogan et al. (49)	+	+	–	–	–	–	–
Rogan et al. (19)	+	+	–	–	–	–	+
Rogan et al. (36)	+	+	–	–	+	+	–
Turbanski et al. (4)	?	?	?	?	–	+	–

Bias rating: (+) = there is a small risk of bias; (−) = there is a risk of bias; (?) = unclear bias because not enough information available.

regions (65). It can be speculated that mechanosensory receptor stimulation can increase cognitive function. Furthermore, it has been assumed that improvement in cognitive function depends on increased production of neurotrophins [e.g., brain-derived neurotrophic factor (BDNF)] (67). BDNF is recognized as the most significant neurotrophic growth factor related to neuronal plasticity and has a key role in the differentiation and survival of neurons (68). Studies could demonstrate a close correlation between increased BDNF levels and WBV (69, 70). However, so far it is unclear how mechanical vibrations may influence the expression of BDNF (71).

The loss of balance ability is an important risk factor for falls in elderly individuals. Reactive balance is a crucial part of avoiding and adapting to complex environments that threaten postural stability (72). In German-speaking countries, the balance ability is considered to be a coordinative ability (73). We describe this ability as the aggregate understood to maintain and regain balance, taking into account the necessary personal conditions.

Various types of exercises (e.g., airex pad, tilting board, swinging platforms) are used in treatment settings and summarized with the synonym balance training (74–77). The goal is to optimize balance. It is assumed that balance is a skill, and that balance training improves several balance tasks at the same time (29). However, recent studies indicate that only those balance tasks that are trained can also improve (77, 78). Giboin et al. (32) showed that the group which trained in a single-leg stand on the tilting board and the group that trained in a single-leg stand on the swinging board (Posturomed) improved statistically significantly only in the area in which they trained.

Recently, there have been attempts to move away from the term ability towards the definition of skill (73). Taube (78) explained that balance training does not change the behavior of the spinal reflex *per se*. It seems rather to improve the ability of finding the right reflex settings for specific conditions of postural control. Thus, balance training improves task-specific reflex modulation. Low et al. (55) postulated, that specific balance exercise could be the only one likely to improve postural balance. Slackline training improves postural balance in young and elderly individuals in a one-leg stance (56, 57). However, the impact of slackline training is limited or negligible for standard static and dynamic bipedal stances (58–60). Paillard (79) explains that specific balance training optimizes postural skills, but it is not known whether these skills improve motor skills in all types of physical activity. He further refers to the fact that additional studies are required to address this question accurately. Grabiner et al. (80) indicates that task-specific perturbation training is superior to traditional balance exercise training in improving reactive balance capacity and thus preventing falls. Kim et al. (72) performed a network meta-analysis to specify which exercise method is most effective to improve reactive balance in elderly individuals. They analyzed data of 39 RCTs including 1,388 elderly individuals receiving balance training with reactive components (perturbations training) demonstrated the most amount of improvement in reactive training, followed by power training and gait training. SR-WBV is power training. SR-WBV has the potential to improve race of force development after four weeks SR-WBV training in elderly individuals (48). In relation to gait, SR-WBV can be used as skilling up in elderly not able to perform standard gait training. It is known that SR-WBV could significantly improve gait in older adults (19, 47, 49). In the case of a reactive balance, the better the gait, the sooner gait training can be started.

Limitation

The observed heterogeneity in the individual studies' study quality and findings impede a clear-cut answer to the research question. Furthermore, the study design of the pilot study does not allow a clear conclusion on efficacy because the primary aim of the pilot study is not to assess an exact intervention effect size, but rather to determine the sample sizes and evaluate feasibility of the study protocol (81–83).

Conclusion

We found a heterogeneous situation on effects for balance according to SR-WBV. One study showed clinical relevance for ETGUG. Two studies examined the skill in reactive balance. Since balance is a skill and SR-WBV trains reactive balance, future studies should focus on the parameter reactive balance.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material, further inquiries can be directed to the corresponding author/s.

Author contributions

This research project was developed by SR. Data collection was undertaken by two independent bachelor students. SR wrote and JT edited the manuscript. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Content validity, interpretability, and internal consistency of the "Quality First" assessment to evaluate movement quality in hop tests following ACL rehabilitation. A cross-sectional study

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Introduction: Current approaches fail to adequately identify sport readiness after anterior cruciate ligament (ACL) rehabilitation. Altered landing biomechanics after ACL reconstruction are associated with increased risk of a noncontact ACL reinjury. There is a lack of objective factors to screen for deficient movement patterns. Therefore, the aim of this study was to investigate content validity, interpretability, and internal consistency for the newly developed "Quality First" assessment to evaluate movement quality during hop tests in patients after ACL rehabilitation.
Method: Participants in this cross-sectional study were recruited in collaboration with the Altius Swiss Sportmed Center in Rheinfelden, Switzerland. After a successful ACL reconstruction, the movement quality of 50 hop test batteries was evaluated between 6 and 24 months postoperatively with the "Quality First" assessment. Content validity was assessed from the perspective of professionals. To check the interpretability, classical test theory was employed. Cronbach's α was calculated to evaluate internal consistency.
Results: Content validity resulted in the inclusion of three different hop tests (single-leg hop for distance, vertical hop, and side hop). The "Quality First" assessment is enabled to evaluate movement quality in the sagittal, vertical, and the transversal plane. After the exclusion process, the "Quality First" assessment was free from floor and ceiling effects and obtained a sufficient Cronbach's α. The final version consists of 15 items, rated on a 4-point scale.
Discussion: By means of further validations, the "Quality First" assessment could offer a possibility to evaluate movement quality after ACL rehabilitation during hop tests.

KEYWORDS
anterior cruciate ligament (ACL), movement quality, return to sport (RTS), measurement properties, hop test, assessment

Abbreviations
ACL, anterior cruciate ligament; BMI, body mass index; M, month; m, mean; n, number; RTS, return to sport; SD, standard deviation; SH, side hop; SLHD, single-leg hop for distance; VH, vertical hop; y, year.

1. Introduction

Return to sport (RTS) outcomes following anterior cruciate ligament (ACL) reconstruction are unsatisfactory (1), whereas current approaches fail to adequately identify sport readiness (2). An underestimated factor may be altered landing biomechanics after ACL reconstruction (3–5), which are associated with increased risk of a noncontact ACL reinjury (3, 6). Achieving symmetrical results in the single-leg hop for distance (SLHD) test does not ensure symmetry in kinematic variables (7, 8). Therefore, the evaluation of movement quality should be included besides quantitative and psychological parameters in RTS assessments after ACL reconstruction (1, 9–11).

Three-dimensional motion analysis as the gold standard to capture biomechanical risk factors for a reinjury of the ACL is not feasible for most clinical settings because of financial, spatial, and temporal costs (4, 12). There is a lack of objective factors to screen for deficient movement patterns (4). The Landing Error Scoring System is a simple tool to identify potentially high-risk movement patterns during a bipedal jump-landing task (13–15).

Bipedal assessments may not be sensitive enough to identify asymmetries in lower extremity (16). In addition, single-leg landings are a typical ACL injury mechanism (17). Consequently, single limb tasks such as single-leg hop tests are important to identify limb asymmetries in movement and landing patterns (18).

A recent systematic review did not reveal an effective assessment to evaluate movement quality during hop tests after ACL reconstruction (15). With this background and an in-depth literature search about the biomechanical risk factors for ACL injuries, our work group created the assessment tool "Quality First". With this assessment under examination, the work group expects to reliably identify deficient landing patterns during single-leg hop tests. To account for the variable injury mechanisms the "Quality First" assessment was developed to evaluate movement quality during landings of the SLHD, the vertical hop (VH), and the side hop (SH) test (19).

To ensure that meaningful data can be extracted from the presented assessment during hop tests after ACL reconstruction, the measurement properties of the tool must be established (20). The aim of this study was to investigate content validity, interpretability, and internal consistency for the "Quality First" assessment during hop tests in patients after ACL reconstruction.

2. Methods

2.1. Study design

This study corresponded to a cross-sectional study design (21) and adhered to the STROBE guidelines (22). The movement quality during hop tests of the participants was evaluated at a specific time point with the "Quality First" assessment.

2.2. Participants

Participants in this study were recruited in collaboration with the Altius Swiss Sportmed Center in Rheinfelden, Switzerland. After a successful ACL reconstruction, patients of this clinic routinely attended a RTS test battery between 6 and 24 months postoperatively. Fifty RTS single-leg test batteries from 27 participants were included. Each test leg was evaluated individually. Only participants who were able to hop unilaterally with each leg were included. If the performance of one leg was incorrect, only the test from the correct leg was included. All inclusion and exclusion criteria are shown in **Table 1**.

The patient characteristics are shown in **Table 2**. The age ranged from 14 to 39 years, 40.7% were female, and the average body mass index (BMI) was 23.7. The mean time since surgery was 10.1 months. In 85.2% and 14.8% of the participants, semitendinosus and quadriceps, respectively, were used as autografts. In three cases, the ACL reconstruction was a revision surgery. Some participants had associated injuries on the meniscus ($n = 17$), ligaments ($n = 3$), cartilage ($n = 4$), or others

TABLE 1 Inclusion and exclusion criteria.

Inclusion criteria	Exclusion criteria
• Time from ACL reconstruction 6–24 months • Attend routine RTS test battery in Altius Swiss Sportmed Center	• Single-leg hopping not possible • Incorrect test procedure (based on instruction document (**Supplementary Figure S2**) • Incomplete RTS test (at least one of the three hop tests missing)

ACL, anterior cruciate ligament; RTS, return to sport.

TABLE 2 Patient characteristics.

Characteristics ($n = 27$)	Value
Age, y (m ± SD)	24.2 (±6.5)
Sex, *n* (%)	
Women	11 (40.7)
Men	16 (59.3)
Height, cm (m ± SD)	175.4 (±7.6)
Weight, kg (m ± SD)	73.3 (±12.0)
BMI, kg/m^2 (m ± SD)	23.7 (±2.7)
Injured knee, *n* (%)	
Right	12 (44.4)
Left	15 (55.6)
Time since surgery, M (m ± SD)	10.1 (±3.1)
ACL revision surgery, *n* (%)	3 (11.1)
Type of graft, *n* (%)	
Semitendinosus	23 (85.2)
Quadriceps	4 (14.8)
Associated injuries, *n* (%)	
Meniscus	18 (66.7)
Ligament	3 (11.5)
Cartilage	4 (15.4)
Other	2 (7.7)
Tegner score before injury (m ± SD)	7.5 (±1.0)

ACL, anterior cruciate ligament; BMI, body mass index; M, month; m, mean; *n*, number; SD, standard deviation; y, year.

(femoral notchplasty, $n = 2$). As visible in the mean Tegner score (7.5), participants mostly participated in competitive sports before surgery.

All participants gave their written general consent prior to participation. The study was evaluated by the Regional Ethical Review Board of "Nordwest- und Zentralschweiz" (Project-ID: 2021–01169).

2.3. Data collection

The "Quality First" construct seeks to describe the latent variable "movement quality" in the form of a reflective model and corresponds to a 66-point scale with three subgroups. A 4-point scale was used to represent different characteristics while maintaining a high discriminability within the items (23). In total, 22 items were rated between 0 and 3. Therefore, the total score ranged between 0 and 66 points. A higher number of points indicated a better movement quality. The subgroups consisted of eight items from the SLHD, the same eight items from the VH, and six items from the SH test. The maximal score for the individual subgroups was 24 for the SLHD and the VH test, and 18 for the SH test. The items included general characteristics like shock absorption during landing and joint-specific movement quality parameters of the trunk, hip, knee, and the foot. An example of the knee evaluation was the item "knee alignment", where the movement quality was rated from 0 ("The knee joint is neutrally aligned with the axis during landing") to 3 ("Extreme knee joint valgus"). Full description of the "Quality First" assessment is available in the **Supplementary Figure S1**).

Data collection took place between 2021 and 2022. After participation was confirmed, SLHD, VH, and SH tests were performed with each leg on the same occasion. For SLHD and VH, three test trials were performed. In case the last trial was the best, the test continued until no improvements were made. For conduction of the SLHD test, participants jumped as far as possible on one leg and landed with the same leg. During the VH test, participants jumped as high as possible on one leg and landed on the same leg. For SLHD and VH tests, the hands of the participants were unconstrained. To generate a valid test, the landing position needed to be held stable for at least 2 s. For the SH, participants jumped on one leg from side to side over two 40 cm apart parallel strips. With their hands placed on their hips, participants were required to jump during 30 s as many times as possible over the two strips, without touching the marking line. For the SH test, only one test trial was performed. The test had to be repeated if more than 25% of the jumps were failed attempts. For each of the three hop tests, the uninjured leg was tested first. Two independent sport scientists from the Altius Swiss Sportmed Center were instructed to videotape the tests in a standardized manner with an iPad® Pro 11.0 (Apple Inc., Cupertino, CA, United States) from a frontal view and to label them with the patient ID (**Supplementary Figure S2**). Through the Dartfish®-Application (Dartfish® 360, Dartfish HQ, Lausanne, CH), the anonymized video tapes were recorded with a frame rate of 120 frames per second and a resolution of 1080p (pixels) and then uploaded to a secured cloud system, where the examiners evaluated the movement quality. Three physical therapists independently observed the videos several times and in slow motion without any time limit. They could watch the video as often as necessary, until they finalized their score with the "Quality First" assessment to rate the movement quality. Later, consensus was made by the same physical therapists and used for statistical calculations in this study.

2.4. Data analysis

The COSMIN guidelines were used to evaluate measurement properties of the "Quality First" assessment (20). The present study was a first step in validating the "Quality First" assessment in participants after ACL reconstruction including content validity, interpretability (floor and ceiling effects), and internal consistency.

2.5. Statistical methods

2.5.1. Content validity

Content validity was assessed from the perspective of professionals (20). Three physical therapists who were candidates for a master's degree in sports physiotherapy with 3–4 years of clinical experience in sports physiotherapy created the "Quality First" assessment with current biomechanical risk factors for ACL injuries based on an internal structured literature search in 2020. The relevance of the included hop tests and the individual items were discussed in this work group with a sports scientist who had a doctoral degree. The structured developing process included a first field test with seven physical therapists working in two different outpatient clinics. After a consensus discussion and an improvement procedure, a second field test was conducted by the work group. The resulting version was finalized in a second consensus meeting between the work group and the sports scientist.

2.5.2. Interpretability

To check if all response options were informative, the classical test theory was employed. The distribution of the score at the item level determined to what extent the response options were used and were presented with frequency tables (24). Floor and ceiling effects occurred, if more than 15% of the participants achieved the lowest or highest possible score of a subgroup (24).

2.5.3. Internal consistency

Internal consistency was assessed to show the degree of the interrelatedness among the individual items and their subgroup (20). Cronbach's α (25) was calculated for each subgroup separately to estimate item-specific variance (26). To explore if a subgroup should be excluded from the assessment, a Cronbach's α between 0.7 and 0.95 was considered adequate (27). Furthermore, Spearman's rank correlation between the items and the subgroup was calculated. Items from a subgroup not attaining a Cronbach's α between 0.7 and 0.95 were excluded if (1) the Cronbach's α

value increased when the item was excluded and (2) the item–subtotal correlation was below 0.2 or higher than 0.7 (24). The work group decided about the exclusion of a subgroup with Cronbach's α values between 0.65 and 0.7 and values higher than 0.95 (28). Furthermore, the difficulty and discrimination of the individual items were presented. All statistical analyses were carried out using the program RStudio (Version 4.1.2, License AGPL v3, Boston, MA, United States). Only complete RTS tests were included; therefore, missing data did not have to be addressed.

3. Results

3.1. Participants

In 4 out of the 27 included participants, only one leg (2 affected, 2 unaffected) was evaluated. This was based on incorrect procedure or incomplete test battery of the second leg. The decision to include unaffected legs was based on the opportunity to ensure more heterogeneity within the jump performances.

3.2. Content validity

The work group discussion resulted in the inclusion of three different hop tests [SLHD, VH, and SH (19)]. With those three subgroups, the "Quality First" assessment is enabled to evaluate movement quality in the sagittal, vertical, and transversal planes. Furthermore, the 30 s time duration of the SH test adds the component of fatigue and possible deterioration of movement quality over time. The work group excluded the item "trunk flexion" because of the difficulty to discriminate between trunk and hip flexion in a 2D frontal view. After the first field test, the items "hip tilt", "hip flexion", and "knee flexion" were excluded in the SH test subgroup due to different biomechanics during a frontal plane hop test. The results of the second field test were used to improve the written explanations of the different items (**Supplementary Figure S1**).

3.3. Interpretability

Distribution of the score at the item level is presented with frequency tables (**Table 3**). The mean scores ranged between 1.7 and 2.8. In terms of the subtotal scores, neither floor nor ceiling effects were observed (**Supplementary material Figures S3–S5**).

3.4. Internal consistency

The total Cronbach's α coefficient (0.72) was considered adequate, and therefore, no complete subgroup was excluded (**Table 4**).

Cronbach's α of the subgroup analysis ranged from 0.58 to 0.65, the item difficulty from 0.07 to 0.87, and the item discrimination from 0.04 to 0.66 (**Table 5**). Based on the improvement of the subgroups' Cronbach's α, if an item was deleted, the following

TABLE 3 Distribution of the score at item level.

Subgroup individual items (%)	0	1	2	3	Mean	±SD
VH						
Shock absorption	0	10	44	46	2.4	0.7
Trunk lateral flexion	0	8	72	20	2.1	0.5
Hip rotation	0	26	52	22	2.0	0.7
Hip tilt	2	24	58	16	1.9	0.7
Hip flexion	2	14	34	50	2.3	0.8
Knee alignment	4	24	48	24	1.9	0.8
Knee flexion	0	2	34	64	2.6	0.5
Foot position	0	14	56	30	2.2	0.7
SLHD						
Shock absorption	0	26	58	16	1.9	0.7
Trunk lateral flexion	0	22	66	12	1.9	0.6
Hip rotation	4	18	52	26	2.0	0.8
Hip tilt	2	20	74	04	1.8	0.5
Hip flexion	0	14	34	52	2.4	0.7
Knee alignment	4	22	58	16	1.9	0.7
Knee flexion	0	0	20	80	2.8	0.4
Foot position	0	2	70	28	2.3	0.5
SH						
Shock absorption	0	4	40	56	2.5	0.6
Trunk lateral flexion	0	16	44	40	2.2	0.7
Hip rotation	0	26	70	4	1.8	0.5
Knee alignment	2	36	50	12	1.7	0.7
Foot position	2	40	44	14	1.7	0.7
Fatigue	4	28	44	24	1.9	0.8

SD, standard deviation; SH, side hop; SLHD, single-leg hop for distance; VH, vertical hop.

items were excluded in a first step: "trunk lateral flexion" in the VH and SLHD tests, "hip flexion" and "foot position" in the SLHD test, and "tiredness" in the SH test. In the second step, the items "shock absorption" and "knee flexion" of the SLHD were additionally excluded (**Supplementary Figure S6**).

Spearman's rank correlation at the item–subgroup level ranged from 0.19 to 0.77 (**Supplementary Figures S7–S9**). "Trunk lateral flexion" (item–subgroup correlation: 0.19) was excluded from the SLHD test, and "hip rotation" (item–subgroup correlation: 0.77) was retained due to a sufficient Cronbach's α of the VH test. All other items met a requested item–subtotal correlation (**Supplementary Figures S7–S9**).

3.5. Final version of the "Quality First" assessment

The final version of the "Quality First" assessment consists of the VH ("shock absorption," "hip rotation," "hip tilt," "hip flexion," "knee alignment," "knee flexion," and "foot position"), the SLHD ("hip rotation," "hip tilt," and "knee alignment"), and the SH ("shock absorption," "trunk lateral flexion," "hip rotation," "knee alignment," and "foot position"). Cronbach's α from 0.64 to 0.66 was considered sufficient after a work group consensus discussion (VH, SLHD, and SH; **Table 6**). The possible subgroup scores of the final 45-point scale version are

TABLE 4 Internal consistency of the "Quality First" assessment.

Subgroup	Mean	±SD	Subgroup difficulty	Subgroup discrimination	Cronbach's α	α if deleted
VH	17.3	2.9	0.75	0.56	0.72	0.61
SLHD	16.9	2.4	0.80	0.47		0.70
SH	11.8	2.4	0.74	0.59		0.56

SD, standard deviation; VH, vertical hop; SLHD, single-leg hop for distance; SH, side hop.

TABLE 5 Subgroup analysis (items to be deleted in italic).

Subgroup	Item difficulty	Item discrimination	Cronbach's α	α if deleted
VH			0.65	
Shock absorption	0.79	0.26		0.63
Trunk lateral flexion	0.71	0.09		0.66
Hip rotation	0.65	0.66		0.52
Hip tilt	0.63	0.42		0.59
Hip flexion	0.77	0.31		0.62
Knee alignment	0.64	0.42		0.59
Knee flexion	0.87	0.29		0.63
Foot position	0.72	0.25		0.63
SLHD			0.58	
Shock absorption	0.37	0.29		0.54
Trunk lateral flexion	0.37	0.04		0.61
Hip rotation	0.67	0.46		0.47
Hip tilt	0.60	0.40		0.51
Hip flexion	0.21	0.11		0.60
Knee alignment	0.62	0.55		0.44
Knee flexion	0.07	0.36		0.53
Foot position	0.75	0.11		0.59
SH			0.63	
Shock absorption	0.84	0.39		0.59
Trunk lateral flexion	0.75	0.37		0.59
Hip rotation	0.59	0.50		0.56
Knee alignment	0.57	0.50		0.54
Foot position	0.57	0.27		0.63
Fatigue	0.63	0.26		0.64

VH, vertical hop; SLHD, single-leg hop for distance; SH, side hop.

21 for the VH, 9 for the SLHD, and 15 for the SH test (**Supplementary Figure S10**).

4. Discussion

This study aimed to investigate the measurement properties of the "Quality First" assessment during hop tests in patients after ACL reconstruction. The final version was free from floor and ceiling effects, obtained a sufficient Cronbach's α, and showed adequate item–subgroup correlations.

In the process to determine content validity, every item was selected due to its relevance as a risk factor for an ACL reinjury. This process was led through a structured literature research, clinical expertise, and discussions in the work group with three physical therapists and a movement scientist. Content validity has been investigated in a similar way (28, 29). In one study, four clinicians were asked to rank five functional tasks about their usefulness regarding the degree of knee flexion (29). In another study, a work group of three clinicians and a focus group including two more experts discussed and determined tasks and postural orientation errors based on current scientific knowledge and clinical experience (28). Based on the reported results, the presented procedure seems to be reasonable to establish sufficient content validity for the "Quality First" assessment.

In terms of the interpretability, neither floor nor ceiling effects were observed in the subgroup scores. Regarding the floor effects, it must be considered that only patients with the ability to perform all three hop tests in a safe execution were included in this study. If a patient is not able to perform hop tests in all movement directions, return to sport clearance should not be considered. A comparable study with a similar scale detected several floor effects using a different definition (>70%) on the item level (28). In the present study, floor and ceiling effects were calculated on the subgroup level, because movement quality consists of the interaction of each item. Each item is meant to discriminate between the performance of one aspect in movement

TABLE 6 Final version of the "Quality First" assessment.

Subgroup	Item discrimination	Cronbach's α	α if deleted
VH		0.66	
Shock absorption	0.27		0.66
Hip rotation	0.64		0.54
Hip tilt	0.38		0.62
Hip flexion	0.37		0.63
Knee alignment	0.41		0.62
Knee flexion	0.30		0.65
Foot position	0.25		0.66
SLHD		0.78	
Hip rotation	0.66		0.65
Hip tilt	0.42		0.88
Knee alignment	0.83		0.43
SH		0.64	
Shock absorption	0.29		0.63
Trunk lateral flexion	0.39		0.60
Hip rotation	0.52		0.55
Knee alignment	0.52		0.53
Foot position	0.32		0.64

VH, vertical hop; SLHD, single-leg hop for distance; SH, side hop.

quality and is allowed to contain floor or ceiling effects due to different item difficulties (24).

Regarding the internal consistency, the subgroups of the final version reached a sufficient Cronbach's α (VH: 0.66, SLHD: 0.78, SH: 0.64). A similar study calculated a comparable Cronbach's α (0.82) for postural orientation errors during the SLHD (28). In a subsequent study, an SH task attained a Cronbach's α of 0.64 for the lateral and 0.82 for the medial landing (30). This SH task was performed in a self-selected pace and with only seven landings instead of the 30 s maximal procedure in this study. The work group of the present study considered the calculated Cronbach's α as adequate due to its dependency upon the number of items in a scale, the homogeneity of the participants, and the fundamental differences between the individual items (26, 31).

The exclusion of the item "trunk lateral flexion" in the VH and SLHD tests can be justified through its unclear role as a risk factor for a second ACL injury. A meta-analysis stated moderate evidence that trunk lateral flexion in SLHD landings after an ACL reconstruction does not differ compared with a healthy control group (7). This item sustained in the final version of the SH may be due to the changes of direction during the test procedure. In this context, an ACL reinjury group showed greater asymmetry of trunk lateral flexion during a change of direction task, compared with a no-reinjury group (11). In comparison with the contralateral leg, another study indicated strong evidence for no difference in peak hip flexion, which could justify the exclusion of the item "hip flexion" in the SLHD test (7). Like the other excluded items, "foot position" in the SLHD showed low item discrimination (0.11) and therefore fails to influence the subgroup score (23). The work group accounted for the exclusion of the item "fatigue" during the SH due to the difficulty in rating this item. Despite this exclusion, the strength of the SH test to evaluate possible deteriorations of movement quality over time sustains in the other five items of this subgroup. The decision to exclude the items "shock absorption" and "knee flexion" from the SLHD in a second step was underpinned due to their negative correlation with the total scale. Two items did not meet the desired item–subgroup Spearman rank correlation between 0.2 and 0.7. The item "trunk lateral flexion" would have been excluded anyways due to an improvement in the subgroup Cronbach's α of the SLHD test, and the item "hip rotation" remained in the final version based on a work group consensus discussion. The conclusion of this discussion was based on the fact that excessive internal hip rotation angles during landings are thought to increase the ACL injury risk (32).

The "Quality First" assessment is the first measurement tool to evaluate movement quality during a hop test battery consisting of SLHD, VH, and SH tests. This tool would allow a clinically friendly ready-to-use approach to include the movement quality in an RTS test after ACL reconstruction. The hop test performance for quantitative measures is simply videotaped with an iPad or a similar device and can be analyzed with a common slow-motion viewer. Before the "Quality First" assessment should be used in clinical practice, further investigations are needed. There is one study in progress that examines the inter- and intrarater reliability and to analyze the correlations between the classic quantitative measures and the movement quality outcomes. Another study under way explores the inter- and intrarater reliability of a real-time execution in contrast to the slow-motion analysis of the "Quality First" assessment.

A current limitation of this study is that the structural validity, construct validity, and responsiveness were not assessed as recommended (20). Those limitations should be addressed in future studies. Another future project is needed to focus on the ability of the "Quality First" assessment to evaluate the risk of a possible reinjury in a long-term follow-up study.

It must be pointed out that the implementation of such a quality control instrument in clinical practice warrants specific training and sound instructions for the users in practice. This is especially important when already test batteries have been applied and those routines have to be changed by adding new and more standardized elements. Due to these issues and the inability to perform one of the hop tasks, an unknown number of participants from the routine sample of patients were excluded.

The "Quality First" assessment could offer a possibility to evaluate movement quality after ACL rehabilitation to guide RTS decisions together with a combination of strength, and other performance measures, psychological readiness, and nonphysical factors. By means of further validations, this assessment could be used to provide the patient with important feedback regarding their readiness to RTS. Additionally, the "Quality First" assessment could lead to tailor specific interventions based on detected deficiencies of the individual patient after ACL rehabilitation.

Data availability statement

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by Ethikkommision Nordwest- und Zentral-Schweiz, EKNZ. Written informed consent to participate in this study was provided by the participants' legal guardian/next of kin.

Author contributions

MM-K: concept, design, data acquisition and data analysis, discussion of results, and wrote initial manuscript. MM and MW: design, data acquisition and data analysis, and discussion of results. HB: supervised the project, concept, design, and discussion of results. SC and MH: test procedure, data acquisition and data handling, and discussion of results. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2023.1180957/full#supplementary-material.

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Sensor-based augmented visual feedback for coordination training in healthy adults: a scoping review

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Introduction: Recent advances in sensor technology demonstrate the potential to enhance training regimes with sensor-based augmented visual feedback training systems for complex movement tasks in sports. Sensorimotor learning requires feedback that guides the learning process towards an optimal solution for the task to be learned, while considering relevant aspects of the individual control system—a process that can be summarized as learning or improving coordination. Sensorimotor learning can be fostered significantly by coaches or therapists providing additional external feedback, which can be incorporated very effectively into the sensorimotor learning process when chosen carefully and administered well. Sensor technology can complement existing measures and therefore improve the feedback provided by the coach or therapist. Ultimately, this sensor technology constitutes a means for autonomous training by giving augmented feedback based on physiological, kinetic, or kinematic data, both in real-time and after training. This requires that the key aspects of feedback administration that prevent excessive guidance can also be successfully automated and incorporated into such electronic devices.

Methods: After setting the stage from a computational perspective on motor control and learning, we provided a scoping review of the findings on sensor-based augmented visual feedback in complex sensorimotor tasks occurring in sports-related settings. To increase homogeneity and comparability of the results, we excluded studies focusing on modalities other than visual feedback and employed strict inclusion criteria regarding movement task complexity and health status of participants.

Results: We reviewed 26 studies that investigated visual feedback in training regimes involving healthy adults aged 18-65. We extracted relevant data regarding the chosen feedback and intervention designs, measured outcomes, and summarized recommendations from the literature.

Discussion: Based on these findings and the theoretical background on motor learning, we compiled a set of considerations and recommendations for the development and evaluation of future sensor-based augmented feedback systems in the interim. However, high heterogeneity and high risk of bias prevent a meaningful statistical synthesis for an evidence-based feedback design guidance. Stronger study design and reporting guidelines are necessary for future research in the context of complex skill acquisition.

KEYWORDS

sensor-based, augmented feedback, visual feedback, motor learning, coordination training, autonomous training, design guideline

Introduction

In the last decades, technological progress has brought about a multitude of competitively priced sensor devices for recording and analyzing human movement in real time. In the context of sports and exercise, this development led to a variety of commercial products leveraging sensor-based augmented feedback applied in domains ranging from physical activity monitoring to classical strength and endurance training to exergaming and even motor-skill learning (1). Such autonomous technological solutions promise to be an efficient and (cost-)effective complement to classical instructor-led interventions and are therefore marketed aggressively for home training, but also for fitness centers and even for clinical use in physical therapy and rehabilitation. The prevalence of human trainers and their obvious benefits in all kinds of sport training alone form strong indicators that such sensor-based augmented feedback training (SAFT) systems may also provide advantages in the aforementioned domains while tackling already prevailing and in the future intensifying cost and personnel capacity issues. Therefore, further investigation of potential benefits but also harms of sensor-based augmented feedback seems necessary.

In general, SAFT systems are intended to foster sensorimotor learning, a process which brings about a relatively permanent improvement in the capability of a person to perform a sensorimotor skill (2). From a theoretical perspective on motor control and learning, four principal sensorimotor learning mechanisms can be distinguished, which extend Newell's well-known task-space landscape metaphor (3) and were first elaborated by Hossner, Kredel, and Franklin (4)—namely, task-space formation, differentiation, exploration and (de-) composition. It quickly becomes apparent that SAFT systems can foster sensorimotor learning during all these stages. First, SAFT systems can assist novices during task-space formation, where learners need to identify basic functional task structures. As Hossner and Zahno (5) state, this process can be enhanced by (i) providing task-goal related instructions, (ii) following appropriate schedules, or (iii) introducing part-whole training. Not only can SAFT systems provide this information in a reliable and systematic manner, moreover, they can analyze the learner's compliance based on the gathered sensor data and adapt to potential deviations. Second, during task-space differentiation, learners start paying attention to less salient task parameters, thus increasing the dimensionality of the task-space. SAFT systems can support this process by inducing controlled amounts of variance, e.g., by increasing difficulty or augmenting errors. This contributes to optimally structured learning contexts that promote the identification of additional task-relevant control variables while, at the same time, assuring the exploration of the continuously evolving task subspaces. Third, SAFT systems allow to point the learner towards better task solutions during task-space exploration and therewith promote a systematic escape from local optima. According to Hossner, Kredel, and Franklin (4), this can be achieved by avoiding repetitive, blocked practice of task variants, which fosters a stronger representation in memory [cf. the reconstruction hypothesis (6)] and facilitates an interpolation of the explored support points of the task space [cf. the elaboration hypothesis (7)]. Fourth, such a well-explored task space can be expected to allow for a better transfer of sub-spaces containing movement structures into the context of different tasks. Consequently, during task-space (de-)composition, learners need to be supported in identifying functional (sub-)structures in their task spaces that can be potentially applied outside the current motor task (5). Applying the above reasoning again, as SAFT systems allow for a systematic variation of specific, functionally relevant control variables while keeping others constant, their application can promote structure detection and therefore (sub-)space identification. Moreover, decomposing a task into such transferrable sub-structures may allow to train those in isolation, increasing the quality of the building blocks independent from training the whole task (4). Functionally relevant task-space decomposition would additionally allow to start task-space exploration with a well-educated guess, consequently changing the learning of completely novel tasks to a transfer of functionally fitting subspaces from previous experience (5). With its fine granularity on sensory motor learning mechanisms, this theoretical framework has the potential to guide the conceptual design of SAFT systems to ultimately benefit sensorimotor learning.

Despite all potential benefits, a major challenge remains for a successful application of SAFT systems to sensorimotor learning: Finding appropriate approaches to guide the learner to specific regions of the task space, in other words, defining the optimal type and amount of instruction and feedback for the current experience level of the individual learner. Well established approaches in sports practice can be differentiated by the amount of structure provided during the learning process. They form a continuum between unsupervised and supervised learning regimes.

On one end of the continuum, and like unsupervised learning, (unguided) discovery learning builds on the self-organized search behavior by the learners, assuming that they can find their optimal task solution better than any external observer [e.g., Vereijken and Whiting (8)]. When targeting specific mechanisms of motor learning as sketched above, this approach seems particularly suited to exploit inherent variability, while a systematic addressing of specific regions of the task space seems limited.

Applying a rather prescriptive approach, located at the other end of the continuum, those specific regions might be targeted more easily by explicitly instructing the learner, ideally in the form of desired sensory consequences. Those instructions are thought to generate sensorimotor imagery together with the desired action consequences and therefore provide sufficient input to the motor system to parametrize the movement (4). While older research found larger detrimental effects due to raised psychological demands for explicitly learned skills (9), in a recent review, Kal et al. (10) did not find clear disadvantages in their descriptive synthesis. They therefore explicitly encourage employing both approaches in practice based on their appropriateness for the task and learning challenge at hand. Nevertheless, applying instructions and feedback excessively may introduce artificial feedback-specific dimensions to the task space which provide highly precise information for movement parametrization. From a Bayesian perspective, the estimations throughout the learning process would be dominated by those artificial dimensions over noisier, task-relevant dimensions. However as soon as feedback is removed, the artificial dimensions do not provide meaningful information anymore, preventing the sensorimotor system from finding a good solution. This phenomenon is known as the guidance effect (11, 12). Even if this effect does not necessarily generalize to more complex tasks [e.g., (13–16)], considering the general mechanism seems sensible.

In their 2002 review, Wulf and Shea (14) concluded that principles derived from simple skill learning do not necessarily generalize and more intensive research on complex skills is required to advance motor learning theory and to adequately inform practice. Since then, most research has been investigating augmented feedback very broadly [cf. Sigrist et al. (17)]. Neglecting given instructions and experience levels while including multiple modalities, mixed populations, and simpler movement tasks in medical settings generally results in a very heterogenous set of outcomes not allowing for a clear-cut synthesis of the results. The combination of these factors may have contributed to the ambiguous result patterns in prior research on augmented feedback in motor skill learning.

In this review, an approach involving a restrictive search purview has been employed to increase the homogeneity of the included research. Diminished health, older age, or different levels of motor development may affect motor learning and the optimality of developed strategies, so we restricted target population to healthy, non-elderly adults. When it comes to the task complexity-dependent effect of feedback, it is still unclear whether it should be regarded as a binary question of simple movements vs. complex movements, or rather as a spectrum. We thus opted for a conservative definition of complex movements that involves postural control and multi-joint movements, further limiting the considered experiments to sports-related coordination training interventions with such complex movement tasks. A previous review on the potential impact of different feedback modalities and parameters has concluded that vision was the most investigated modality (17), which can be enforced from an implementational viewpoint due to the ubiquity of electronic screens in digital technologies and existing training devices. By focusing on visual feedback as the largest body of evidence only, we expect to maximize the review's synthesis potential. To sum up, the objective of our scoping literature review is thus to provide the basis for informed feedback design and to provide guidelines for the development of future autonomous visual SAFT systems for sports-related settings to maximize the training benefits derived from such feedback. More specifically, we approach this objective by addressing the following goals:

i. Aggregate results pertaining to similar feedback regimes to provide an overview of the findings in relation to these choices.
ii. Outline what visual feedback regimes have been considered in sports-related research.
iii. Compile the recommendations made in these studies regarding visual feedback regimes.

Methods

We followed the PRISMA Extension for Scoping Reviews (PRISMA-ScR) (18) without prior registration of a formal review protocol. A research librarian advised the investigators in the selection of the databases and the formulation of the search strings. In accordance with the recommendations of the Interim Guidance from the Cochrane Rapid Reviews Methods Group (19), the three electronic databases Embase, PubMed, and Cochrane Central were searched to cover a comprehensive basis of the available literature. The last search on each database was carried out on the 17th of October 2022 by one investigator. The search strings consisted of a conjunction of disjunctions, grouped into the following four inclusion criteria (with *NEAR/10* meaning that the respective keywords need to be closer than ten words):

- **Feedback:** *("knowledge of performance" OR "knowledge of results" OR ((augment* OR external OR extrinsic OR kinetic* OR kinematic* OR motion) NEAR/10 (feedback OR biofeedback)))*
- **Coordination:** *(performance OR motor OR movement OR skill* OR coordination OR neuromuscular OR techni* OR athlet* OR sport*)*
- **Training:** *(training OR acquisition OR improvement OR learning OR athlet* OR sport*)*
- **Visual:** *(visual* OR display* OR screen OR perceptual*)*

The search was limited to articles published in peer-reviewed journals and always covered abstracts. If the database interface permitted a combined search with titles and keywords, then these were also included. Where possible, filters were set to exclude reviews and study registrations and to only consider intervention studies. If this was not possible, the filtering process was performed manually in the screening phase. There was no restriction to sensor-based feedback in the search terms because such specifics of the methodology may be missing in the abstract.

The screening procedure consisted of two phases: The first phase was based on abstracts, titles, and keywords, while the second phase considered the full-text articles. In both phases, two screeners read all records. After the first phase, 52 items had conflicting verdicts, which were then discussed on a one-by-one basis until a consensus was reached between both screeners. After the second phase, all results were discussed to verify the final selection. Studies in languages other than English were excluded, as well as studies older than 30 years (publication year 1991 or earlier) as sensor-based real-time feedback was practically unavailable before. Studies were excluded if they did not include a complex sports-related coordination task with sensor-based visual feedback or did not have at least one group of healthy, non-elderly adult participants. The general rationale behind these criteria was mostly based on the theoretical aspects that were discussed in the introduction. A practical explanation with the resulting concrete differentiations in the screening procedure is given here:

- Sensor-based feedback:

  Our goal was to restrict the purview to feedback that was generated in an automatic and objective manner, as opposed to, e.g., human augmented feedback from coaches or peers. This decision has some unintuitive consequences: Video-feedback was included, because it is technically a sensor, while other visual feedback generated by electronic devices such as laser pointers was not included.

- Visual feedback:
  By focusing on one feedback modality, we hope to attain more consistent results. However, we still included studies that added other feedback modalities to the provided visual feedback if the visual feedback was clearly in the focus. Other intervention groups with different feedback modalities or no feedback at all were considered as control groups for the data extraction.
- Healthy, adult, non-elderly population:
  Disorders, diseases, and age could affect motor learning mechanisms, because these factors might alter the optimality of specific movement solutions and because cognitive maturity or decline might affect motor learning. Thus, as a rather conservative boundary, we only considered participants that are between 18 and 65 years. If a study involved at least one group of participants that fully satisfies these criteria, then the study was included even if other groups were considered in the study. In that case, all groups not satisfying these criteria were ignored during the data extraction.
- Sports-related, complex sensorimotor tasks:
  We expected participants to have a different mindset in sports-related training compared to medical settings. Compared to sports, interventions targeting activities of daily living (ADL) generally have a different focus, and, in turn, a potentially different feedback objective. Therefore, we excluded ADL and simple balancing tasks.

  We purposefully drew the line between simple and complex tasks rather conservatively so that any study lying between clearly complex and clearly simple tasks was excluded as well. This should ensure that possible negative outcomes stemming from tasks that were not quite complex enough are fully avoided in the synthesis of outcomes, but it is in no way meant as a definition for what constitutes a complex movement task. Tasks which required active control of only one single joint were excluded, as well as bimanual tasks such as reaching, pointing, or sequencing. On the other hand, rowing studies were included despite the seated position if the correct execution of the task required coordination of leg, hip, and trunk movements in addition to the movement of the arms.

After the full-text screening, included studies were categorized into three distinct groups according to the applicability of their results for a potential synthesis. First, if a study reported on the difference between pre- and post-tests for intervention and comparable control groups, with all participants satisfying our population inclusion criteria, then it was categorized as reporting a *training effect*. This category has the potential to indicate how visual feedback design affect retention effects.

For the control group to be considered as comparable, we required that it was different from the intervention group, both regarding participants (i.e., a distinct set of people) and the provided feedback: the control must have either no feedback, a different feedback modality, or also visual feedback but with a relevant change to the way it is designed or administered. Furthermore, the feedback must be withdrawn during testing for all groups to ensure that the measured effects stem from changes in the motor skill in the original task. The measured effect must therefore constitute actual learning and not just a temporary effect caused by the task difference brought about by the given feedback. Second, a study that compares feedback trials with no-feedback trials was categorized as reporting *immediate effect* of feedback. The control can again consist of no-feedback, a different modality, or visual feedback with some aspects changed. Contrary to the first category, these studies must necessarily include tests or measurements with feedback. The control group can either be a different group of participants like in the first category, or alternatively the same group under different feedback conditions in a within-subject design. Therefore, whereas the first category required at least two groups of participants satisfying our population inclusion criteria, one such group was enough to categorize the study as reporting on immediate effects. Third, all other studies were only deemed relevant from a *design-only* perspective, with the focus on the design choices rather than their results. To be included in this category, studies still had to satisfy our inclusion criteria, but they either had exactly one participant group satisfying our population criteria and no within-subject design, or they had multiple participant groups that were not comparable because they did not differ in the administration of the visual feedback (for example only differing in other feedback modalities administered in conjunction with visual feedback).

For the structured data extraction, two investigators extracted information and co-edited the results into a table. Conflicting table entries were discussed until a consensus was reached. The table was then stratified so that all entries follow common nomenclature, and further condensed into the two final, more concise tables presented in this article. The study characteristics were summarized in a first table (**Table 1**), where the columns broadly describe the category, the task and its goal, the intervention, and the participants for each study. A second table (**Table 2**) was split into the three study categories (training effect, immediate effect, design-only) by horizontal lines, using multiple rows for reports including multiple studies, depicting details of the outcomes and the visual feedback regimes for each study. For each main outcome of the studies in the training effect category, at most one post-test (PT) directly following the last intervention session, one short-term retention test (RT1) at least 1 day after the last intervention session, and one long-term retention test (RT2) were considered, each of which is represented in a different column. Potential additional retention tests were discarded because they would only describe the pattern of depreciation over time in more detail. Since the time effect of the interventions in these studies cannot be clearly separated from the immediate effect of the feedback, measurements during the intervention phase were not considered for this study category. Conversely, such immediate tests (IT) were considered

TABLE 1 Overview of tasks, goals, interventions, and population characteristics.

Identifier	Type	Task	Goal	Duration	Sessions	Groups	*N*	Age	Sex	Experience
Benjaminse et al. (20)	TE	Sidestep	Reduce peak knee forces	1	1	3	90	24.6 ± 4.4*	X	Advanced
Chan et al. (21)	TE	Treadmill Running	Soften footfalls	14	8	2	320	18–50	X	Intermediate*
Ericksen et al. (22)	TE	Jumping	Stick the landing	1	1	3	36	20.7 ± 2.3*	F	Beginner
Gilgen-Ammann et al. (23)	TE	Running	Reduce ground contact time	28	8	3	30	31.0 ± 7.5	X	Advanced
Mononen et al. (24)	TE	Shooting	Maximize accuracy	28	12	4	34	20.4 ± 1.8	M	Intermediate
Mulloy et al. (25)	TE	Fencing Lunge	Maximize propulsion, keep sequencing	180	6	2	32	18–40	X	Novice
Nagata et al. (26)	TE	Jump Squats	Increase lifting velocity	28	7	4	37	19–22	M	Advanced
Nekar et al. (27)	TE	Squats	Maintain proper form	28	12	4	48	18–35	M	Beginner
Post et al. (28)	TE	Golf Chipping	Hit target, maintain form	1	1	2	44	21.8 ± 1.3	X	Novice
Rauter et al. (29)	TE	Rowing	Follow reference	2	2	5	40	19–32	X	Novice
Rauter et al. (30)	TE	Rowing	Match target movement	2	2	2	16	27.7 ± 1.9	X	Novice
Rucci and Tomporowski (31)	TE	Hang Power Clean	Maximize power output	28	7	3	17	18–22	F	Intermediate
Sigrist et al. (32)	TE	Rowing	Match target movement	3	3	4	35	28 ± 3.7	X	Novice
Todorov et al. S1 (33)	TE	Table Tennis Return	Hit target through barrier	1	1	3	42	NA	X	Novice
Todorov et al. S2 (33)	TE	Table Tennis Return	Hit target through barrier	3	3	2	18	NA	X	Novice
Viitasalo et al. (34)	TE	Shooting	Maximize accuracy	84	36	4	30	37.5 ± 11.3*	M	Beginner
Anson et al. (35)	IE	Treadmill Walking	Reduce trunk variability	1	1	1*	10*	22.6 ± 4.9	X	Intermediate
Eriksson et al. (36)	IE	Treadmill Running	Adjust running technique	1	1	1	20	28.4 ± 6.4	X	Advanced
Hamacher et al. (37)	IE	Walking	Achieve a balanced gait in frontal plane	1	1	1*	15*	45–65	F	Intermediate
Jones et al. (38)	IE	Ergometer Cycling	Increase performance	21	4	2	20	35.5 ± 6.5*	M	Advanced
Koritnik et al. (39)	IE	Stepping	Match reference	1	1	2	23	23–30	X	Intermediate
Washabaugh et al. (40)	IE	Treadmill Walking	Use full range of motion of knee joint	1	1	1	13	21.0 ± 2.5	X	Intermediate
Weakley et al. (41)	IE	Back Squat	Maximize concentric power	14	4	1	12	21.8 ± 0.9	M	Intermediate
Sigrist et al. (42)	DO	Rowing	Match target movement	2	2	3	24	26.1 ± 3.0	X	Novice
Teng et al. (43)	DO	Treadmill Running	Increase trunk flexion	28	4	1	12	23.3 ± 3.8	X	Intermediate
Teran-Yengle et al. (44)	DO	Treadmill Walking	Avoid knee hyper-extension	1	1	1	17	26.6 ± 5	F	Intermediate

The studies are specified by category (type: TE, training effect; IE, immediate effect; DO, design-only), task, goal, characteristics of the intervention (duration in days, sessions, groups), and population: *N* = number of participants, age (years, either as range or as $M \pm SD$), sex (M, male; F, female; X, mixed), and experience. NA means not available.
*Adjusted by review authors (only counting healthy, adult, and not elderly participant groups; aggregated age; different definitions for experience levels).

for the studies in the immediate effect category, where the focus is not on the effect of the intervention over time but rather on how the feedback affects performance at the instant when it is applied. Finally, no outcome measures were reported for the design-only category because these studies are only relevant for the overview of feedback regimes in the literature, i.e., goal (ii) of this review. The outcomes were represented by arrows indicating whether participants in the visual feedback intervention performed significantly better (⤊), significantly worse (⤋), or not significantly different (⇔) when compared to the control group. For the training effect category, these reported effects always refer to the learning rates or the change from baseline to post- or retention tests (PT, RT1, RT2), in other words group-by-time interaction effects. Conversely, immediate effect studies always refer to the group effects measured (IT), while potential time effects were discarded. Other tests in the respective categories were not reported in the table. In case of differing outcomes, effects for multiple main outcomes were represented separately by splitting them into multiple lines while comparisons to multiple control groups were separated by commas. Multiple visual intervention groups were addressed by prefixing these comparisons with a letter assigned to the different groups (for more details on the chosen nomenclature, refer to the note below Table 2). The chosen intervention groups could have multimodal feedback, but visual-only groups were preferred if available, in which case additional multimodal groups would be disregarded in the reporting of outcomes.

Study populations were classified according to our estimation of their experience in performing the specific movement task. This classification does not necessarily coincide with the one used in the corresponding reports, which were usually based on levels of competition of the recruited participants instead. We classified participants as *Novice* if they had likely no prior experience with the task. Further, *Beginner*, *Intermediate*, and *Advanced* refer to some experience, regular experience, and expert-level experience with the task, respectively.

The qualitative extraction of the recommendations made in the literature was a less structured process. The discussion and conclusion sections of the included studies were screened for statements that we deemed relevant and generalizable for informing future feedback design. Such statements were only extracted if they satisfied two additional conditions: they were based on the results found in the study (as opposed to other referenced research), and they went beyond descriptions and explanations of the outcomes. Two reviewers marked potential candidate passages in the text, and one reviewer then made the decision whether they should be picked up in the result section of this review. The intention was to include only the most important statements in a concise overview.

Finally, one investigator performed a risk of bias assessment using the risk-of-bias tool for randomized trials (ROB 2) (45) for each study in the training effect category. The rationale for this assessment was to evaluate the strength of evidence that a potential meta-analysis could provide in a systematic review of this research topic.

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TABLE 2 Overview of dependent variables, applicable effects, and feedback regimes.

Identifier	CG	Outcome Measures	IT	PT	RT1	RT2	Feedback Measures	Content	KP	KR	C	T	R	F [%]
Benjaminse et al. (20)	Coach, No	Segment Angle (Trunk) Biomechanical Measures	NA	–	⇑, ⇔ ⇔	–	Scene^{T}	Video	X	–	–	X	X	100
Chan et al. (21)	No	Ground Reaction Force Injury Occurrence	NA	⇑ –	–	– ⇑	Force^{T}	Plot	X	–	X	–	–	≈67^{fH}
Ericksen et al. (22)	Coach^{M}, No	Joint Angles (Hip, Knee) Ground Reaction Force	NA	⇔, ⇑	–	–	Segment Position^{T}	Segments + Line^{M}	X	–	X	X	X	100
Gilgen-Ammann et al. (23)	No	Ground Contact Time	NA	–	⇑	–	Mean Time	Bar + Num.	X	–	–	X	X	100H
Mononen et al. (24)	Visual	Score + Score Variability Directional Errors	NA	–	F: ⇑; P: ⇔ ⇔	⇔	Aiming-Point^{T} & Position + Score	Target + Trace & Target + Num.	X –	– X	–	X	– X	F: 100; P: ≈50 100
Mulloy et al. (25)	No	Angular Velocities (Hip, Knee, Ankle)	NA	–	⇔	–	Maximum Angular Velocity + Timing	Color Bar Chart	X	–	–	X	X	≈70
Nagata et al. (26)	No, Coach	Barbell Velocity	NA	⇔	–	⇔	Scene^{T} Mean Velocity	KP: Video; KR: Num.	X –	– X	–	X	–	100
Nekar et al. (27)	Coach^{M}, Visual, No	Knee Extension + Balance Knee Flexion Flexibility	NA	⇔, ⇑, ⇑ ⇔, ⇔, ⇑ ⇔	–	–	Scene^{T}	AR^{M}	X	–	–	X	X	100
Post et al. (28)	Visual^{Y}	Accuracy Form Score	NA	–	⇔	Transfer: ⇑	Scene^{T}	Slow-Motion Video + Video	X	–	–	X	X	100^{S}
Rauter et al. (29)*	Haptic	Spatial Error Velocity Error	NA	–	⇔	–	Position^{T}	Oar Trace	X X	– –	X –	– X	X X	≈70
Rauter et al. (30)	Visual^{Y}	Spatial Errors	NA	–	⇔	–	Position^{T}	Trace^{M}	X	–	X	–	X	≈70E
Rucci and Tomporowski (31)	Verbal	Strength + Power Form Score	NA	⇔ ⇓	–	–	Scene^{T}	Video	X	–	–	X	–	100
Sigrist et al. (32)	Haptic, Audio	Absolute Angular Error Scaling Error Rotation Error Velocity Error Movement Variability	NA	–	⇔ ⇔ C: ⇔; T: ⇑ C: ⇔; T: ⇔, ⇑ C: ⇔, ⇑; T: ⇔	C: ⇔; T: ⇔, ⇑ ⇔ ⇔, ⇑ ⇔ ⇔	Position^{T} + Orientation^{T}	C: Oar; T: Oar + Trace	X	–	X –	– X	X	C: 72 T: ≈72^{S}
Todorov et al. S1 (33)	Coach^{M}	Accuracy Score	NA	⇑	–	–	Position^{T} & Score	Trace & Num.	X X	– –	X –	– X	X X	100
Todorov et al. S2 (33)	Coach	Accuracy Score Trajectory-Distance Score	NA	⇑	–	–	Position^{T}	Trace & Num.	X	–	X	X	X	≈33
Viitasalo et al. (34)	Visual, Coach^{M}	Accuracy	NA	⇔	–	–	Aiming-Point^{T} & Forces & Scene^{T} & Position + Score	Target + Trace & Num. & Video^{M} & Target + Num.	X X –	– – X	– – –	X X X	– X –	≈14 ≈17 100
Anson et al. (35)	No	Low-Frequency- Translational Variance Various Gait Parameters	⇑ ⇔	NA	NA	NA	Position^{T}	Target + Dot	X	–	X	–	–	100
Eriksson et al. (36)	Audio^{W}	Vertical Displacement Step Frequency	⇔	NA	NA	NA	Positions^{T} + Mechanical Power^{T}	Bar Chart	X	–	X	–	X	100
Hamacher et al. (37)	No^{W}	RoM, Inclination (Pelvis) RoM, Inclination (Trunk)	⇔ ⇑	NA	NA	NA	Segment Angles^{T}	Avatar + Axes	X	–	X	–	X	100
Jones et al. (38)	Visual^{W}, Visual	Time, Speed, Power, Perceptual and Physiological Measures	⇑, ⇔	NA	NA	NA	Position^{T} Distance^{T}	Avatars Num.	X	X	X	–	X	100
Koritnik et al. (39)	Visual	Spatial & Temporal Adaptation	⇑	NA	NA	NA	Joint Angles^{T}	Avatar	X	–	X	–	X	100
Washabaugh et al. (40)	No^{W}	Joint Angle + Aftereffects (Knee), Muscle Activation	⇑	NA	NA	NA	Joint Angles^{T}	Bar Chart	X	–	X	–	X	100

(*continued*)

TABLE 2 Continued

Identifier	CG	Outcome Measures	IT	PT	RT1	RT2	Feedback Measures	Content	KP	KR	C	T	R	F [%]
Weakley et al. (41)	Coach^{W}, Coach^{W}, No^{W}	Barbell Velocity	⇔, ⇔, ⇑	NA	NA	NA	Mean Velocity	Num.	X	–	–	X	–	100
Sigrist et al. (42)*	NA^{W}	Spatial & Temporal Errors (Angle + Angular Velocity)	NA	NA	NA	NA	Position^{T}	Oar^{M} Trace^{M}	X X	– –	X –	– X	X X	≈70^{E}
Teng et al. (43)	NA^{W}	Kinematics (Trunk), Joint Kinetics (Hip, Ankle), Automaticity	NA	NA	NA	NA	Segment Angle^{T} Score	Dots Num.	X X	– –	X –	– X	X –	≈50^{fH}
Teran-Yengle et al. (44)	NA^{W}	Joint Angle (Knee)	NA	NA	NA	NA	Joint Angle^{T}	Plot	X	–	X	–	X	≈50

The studies are ordered by category (first training effect, then immediate effect, then design-only, see Table 1). The 'CG' column specifies the type of control groups (instruction/feedback conditions: No, Visual, Haptic, Audio, Coach, ^{M} = multimodal, ^{Y} = yoked control group, ^{W} = within-subject comparison). Interaction effects of feedback for immediate testing (IT), post-test (PT), first retention test (RT1), and last retention test (RT2) are shown by arrows. ⇔: no significant difference between intervention group (IG) and control group (CG), ⇑: IG significantly better, ⇓: IG significantly worse; if effects are identical for multiple outcome measures and/or control group comparisons, they are summarized by one arrow, in case of differing effects: A comma ',' splits control groups, a semicolon ';' splits feedback groups, letters are assigned if multiple visual feedback regimes were applied (F, full; P, partial; C, concurrent; T, terminal). NA means not applicable or not reported. ^{T} indicates that the feedback measure had a time-component. In addition to feedback measure and graphical content of the visual feedback display (properties are linked by a plus sign '+' if presented simultaneously, while an ampersand '&' links multiple quantities separately shown), the following properties of the feedback regime are shown: knowledge of performance (KP), knowledge of results (KR), concurrent (C) or terminal (T) feedback, whether a reference (R) was available or not (X = applied, - = not used), and feedback frequency (F in %). Frequency is either given as fixed percentage of trials/time, as overall average percentage for fading (^{f}), or as maximum allowed percentage for self-selected (^{S}) and error-based (^{E}) regimes, without considering reported additional home-exercise (^{H}).

*Rauter et al. (29) utilized the same visual feedback group's data as the unimodal visual group in Sigrist et al. (42), so their feedback regime was clearly identical without adding to the available evidence. We instead listed the multimodal audiovisual feedback group for Sigrist et al. (42).

Results

Study selection and data extraction

The initial literature search identified 892 records from three databases (**Figure 1**) (46). After removing 224 duplicates, 668 distinct records remained. From these, we excluded all records that did not satisfy the criteria specified in the methods section: 588 records were excluded in the abstract screening stage and 55 reports during the full-text screening, leading to 25 reports included in the final dataset. 15 of these 25 reports measured training effects of visual feedback, but one report consisted of two empirical studies, so in total 16 studies were assigned to the training effect studies. The remaining 10 reports were not eligible for the training effect category because some only had one intervention group satisfying our population criteria (7 reports), no post-intervention tests without feedback were performed (2 reports), or because the control groups differed in other feedback modalities without affecting attributes of the visual feedback (1 report). Of these 10 reports, seven measured performance under different visual feedback conditions and were thus eligible for the immediate effect category, featuring five within-subject designs, one between-subject design, and one with both within- and between-subject comparisons. The remaining three reports did not compare immediate performance under different visual feedback conditions, but instead reported training effects over time for a single group (2 reports) or had control groups that all received the same visual feedback (1 report). All 26 studies of the 25 reports and their characteristics deemed relevant for this review are summarized in **Table 1** (population and intervention) and **Table 2** (dependent variables and feedback).

Several small adjustments were made during the data extraction process. Two studies incorporated groups of participants that did not match our population criteria (35, 37), these groups were subsequently ignored in the data extraction. Multimodal groups receiving visual feedback were disregarded in three studies (31, 39, 42) in **Table 2** because visual-only intervention groups and non-visual control groups were available. Rauter et al. (29) designated the visual feedback group as control group, but for our purposes this constitutes the intervention group, with the haptic feedback groups serving as control instead. In three studies (23, 24, 34), one "true" control group, in which the participants received no intervention at all, was disregarded in **Table 2**. An item of concern was that Rauter et al. (29) and Sigrist et al. (42) seemed to share the same visual-only feedback group, i.e., only one unique dataset was gathered for both studies. The visual-only feedback group is therefore counted twice in the columns of **Table 1** that concern study participants. This group was assigned to Rauter et al. (29) as the main intervention group in **Table 2** so that it could be counted for group comparisons in the training effect outcomes. Because Sigrist et al. (42) is in the design-only category, the same group is not relevant for group comparisons here, so this group was ignored for this study in **Table 2** to avoid over-representation of the same feedback regime. Instead, the otherwise similar multimodal group was considered as the main intervention group in Sigrist et al. (42).

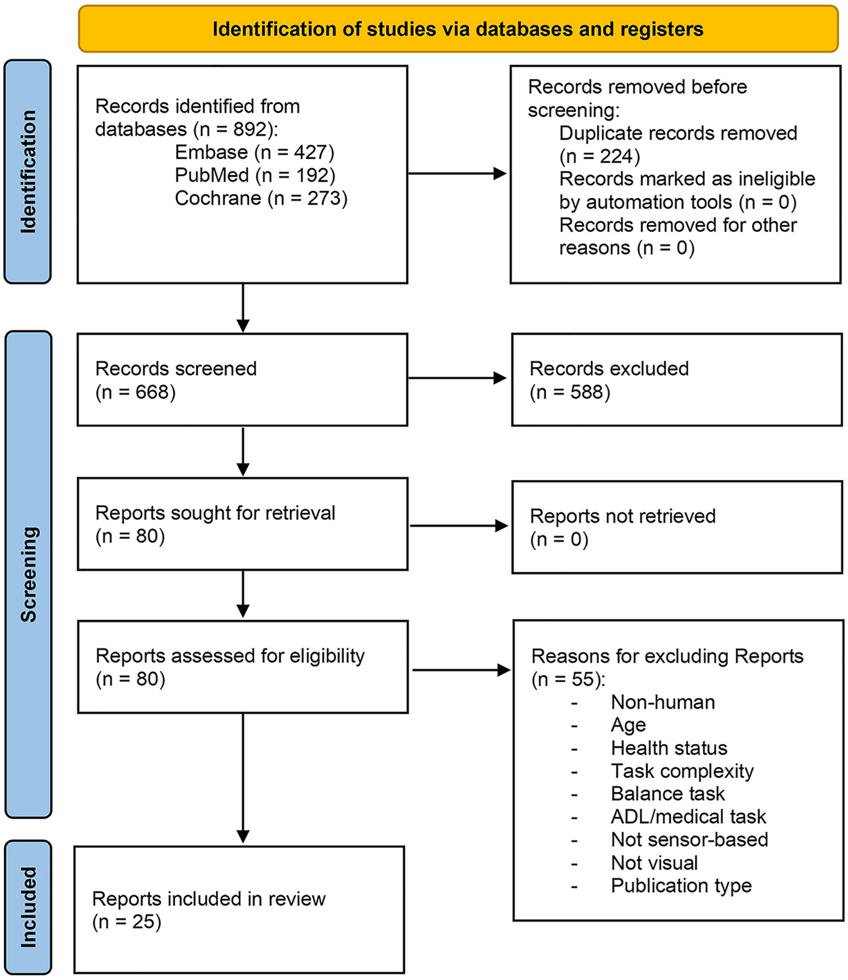

FIGURE 1
PRISMA 2020 (46) flow diagram: overview of the study selection process consisting of database searches, abstract screening, and full-text assessment.

Populations and intervention formats

Most studies (18 out of 26) had a relatively small group size with less than 15 participants per intervention arm (**Table 1** columns "Groups" and "N"). The exceptions were Benjaminse et al. (20) with 30, Chan et al. (21) with 160, Mulloy et al. (25) with 16, Post et al. (28) with 22, Eriksson et al. (36) with 20, Hamacher et al. (37) with 15, and Teran-Yengle et al. (44) with 17.

Instructions were often implicit to the task, e.g., trying to hit a target implicitly conveys the desire to increase accuracy, which was the goal in 5 out of 26 studies. Increasing physiological power output was the objective in 5 studies. More nuanced instructions consisted of following a target movement (5 studies), reducing joint strain (2 studies), or a direct adjustment to the movement technique (11 studies). Two studies explicitly combined the performance goal with the demand to maintain proper technique.

When classifying the studies according to their intervention schedule, 10 studies lasted for less than 1 day, encompassing a single session, while 5 studies lasted between 2 and 3 days with 2–3 sessions. Nine studies lasted between 2 and 4 weeks with 4–12 sessions; the remaining 2 studies lasted 12 weeks with 36 sessions and 6 months with 6 sessions, respectively.

Utilized visual feedback regimes

The quantities used for the feedback mostly consisted of positions, joint angles, or forces relevant to the movement task, often coinciding with one of the dependent variables (cf. "Feedback Measures" and "Outcome Measures" in **Table 2**). These quantities were mostly measured using motion capture systems, cameras, force plates, and inertial measurement units. Todorov et al. (33) used an electromagnetic sensor to track paddle position and orientation. Nekar et al. (27) employed a mobile AR device. The rowing studies (29, 30, 32, 42) all utilized the same rowing simulator, which incorporated rope robots, motion capture, and wire potentiometers. The shooting studies (24, 34) employed an optoelectronic shooting system to detect the shot and to determine the relevant performance metrics. The shooting studies also included a trace of the point where the shooter was aiming at. Nagata et al. (26) used an optical encoder system to measure lifting velocity. In Eriksson et al. (36) and Weakley et al. (41), a position transducer measured the displacements and velocities, respectively. In Jones et al. (38), participants trained on a cycle ergometer. Participants in Washabaugh et al. (40) wore an exoskeleton that measured joint angles (while also applying the resistance for the movement task). Teng et al. (43) included the percentage of time spent in the desired parameter range as terminal feedback in addition to the concurrent joint angles measured by a motion capture system.

Knowledge of Performance (KP) feedback was given in every study, with four studies additionally including Knowledge of Results (KR) in the augmented feedback, but the timing of KP and KR feedback varied between studies. For KP, concurrent and terminal feedback was approximately equally common (in 16 and 17 studies respectively, shown in columns "C" and "T" of **Table 2**). One study, Sigrist et al. (32), reported a deliberate delay of terminal KP feedback during the trials: After feedback was requested by the participant, there was a 10 s delay, after which feedback was shown for the last 18 s of the movement. KR was given as terminal feedback in 3 of 4 studies, with only Jones et al. (38) giving concurrent KR feedback during their trials by displaying the total distance covered.

In 21 studies, some form of reference was incorporated to the visual feedback (as indicated in column "R" of **Table 2**). Possible forms of reference were ideal values or ranges (e.g., given as a line), a virtual avatar or a reference-oar performing the correct movement, or a split-screen video with another performance. Hamacher et al. (37) provided a reference by showing the current joint angles with the desired ranges overlaid on a virtual avatar of the participant. The data for the provided references was either sourced *a priori* (e.g., from recommendations or from experts showing the correct movements) or generated during the study from a participants' previous performances.

According to the following classification into four groups (plots, numerical, video, complex graphics), the 26 studies featured a total of 38 occurrences of graphical feedback visualizations (see column "Content" in **Table 2**). These visualizations varied in terms of graphical complexity and abstraction level, but no study tried to graphically convey more than three quantities at once and no study reported issues with the understandability of the graphics. In 12 studies the feedback was visualized by plotting it on a 2-dimensional plane. This was achieved with linked motion-capture marker-models (1 study), showing the trace of the movement on a plane (5 studies) or in a 3D virtual environment (2 studies), quantity-time plots (2 studies), dots on quantity-quantity plots (1 study), and markings on virtual bulls-eye targets (3 studies, two of which included aiming-traces). In 11 studies, numbers were represented as numerical values or vertical bars. A video recording of the participant was used in 6 studies, one of which involved augmented reality with graphical movement guidance. More complex graphical representations (9 studies) involved virtual avatars, a virtual copy of the training environment to show the trace in, or a virtual rowing simulator that included a virtual representation of the oar and other modalities (e.g., traces). In Jones et al. (38), the avatar was set on a virtual cycling track that graphically simulated a movement through space dependent on their cycling performance. Five studies (22, 27, 30, 34, 42) applied additional non-visual feedback in the visual feedback group, so the participants received multimodal feedback. Audio resulting from the simulation of water in the rowing studies (29, 30, 32, 42) were considered part of the immersion and not specifically marked as multimodal feedback in the table. Analogously, the virtual extension of the oar was not treated as visual feedback. All groups in all rowing studies received this audio and visual feedback.

A form of summary feedback (i.e., feedback that is not specific to a single movement execution) was used in Nagata et al. (26) by averaging over the whole set, and in Gilgen-Ammann et al. (23) by providing only the mean ground contact time over each interval run. Jones et al. (38) was the only instance where participants were deliberately deceived about the nature of the provided feedback: One group was told in one trial that the pacer (the reference avatar) showed their own performance from a baseline trial, without telling them that its speed was increased by 2%.

The reported frequency of each feedback schedule refers to the percentage of trials or time during the intervention phase in which participants had the opportunity to receive feedback (**Table 2** column "F"). Test trials without feedback were treated the same as training trials without feedback if they consisted of the same movements. For the instantaneous effect studies, the frequency was generally 100% because there was no meaningful intervention phase to average over. The only possible exception is Jones et al. (38), which received a + 2% and a + 0% pacer as feedback for 25% of the time each, with the remaining 50% of the total time being reserved for baseline tests without pacer. In 18 studies, the feedback schedule was completely predetermined for at least one visual feedback group. In 8 studies, at least one group received visual feedback with other scheduling strategies. Fading feedback (a gradually decreasing frequency over the intervention duration) was used in Chan et al. (21) and Teng et al. (43). Self-selected feedback (providing feedback only upon request by the participant) was used in Sigrist et al. (32) and Post et al. (28). Self-selection led to variable feedback frequencies considerably different from the maximum possible frequencies, e.g., resulting in a mean frequency of 9% (range 2%–37%) compared to 100% possible in Post et al. (28). Error-based feedback (no or reduced visual feedback when performing below a certain error threshold) was used in three of the four rowing studies (29, 30, 42). Specifically, the trace was only drawn above the error threshold in Rauter et al. (29) and Sigrist et al. (42), and the transparency of the reference oar was increased with decreasing error, making it harder or even impossible to see. In Rauter et al. (30), visual feedback was provided if the spatial error was the dominant error, otherwise an auditive feedback was given for the velocity error instead. Three studies (21, 23, 43) explicitly reported that participants continued training outside the intervention sessions during the intervention period, at home or elsewhere. For these studies, the reported frequencies only refer to the training during the trials, other training (at home without feedback) was not taken into account.

Effect of visual feedback on intervention outcomes

Using a vote counting approach, it is evident that the reported effectiveness of feedback varies a lot between studies (see **Table 2**, where votes are indicated by arrows). When interpreting these outcomes, it is crucial to also consider what exactly the intervention groups were compared against: Even the control groups showed high heterogeneity, which makes a fair comparison impossible. Only one study, Rucci and Tomporowski (31), reported that the visual feedback group showed worse outcomes than their control group, which received verbal feedback. Positive and no benefits are approximately equally common in the feedback and no-feedback conditions of the training effect studies. Even when looking only at the studies with the biggest group-sizes, the outcomes are mixed: Chan et al. (21) (160/group with fading) shows a clear benefit, Benjaminse et al. (20) (30/group with 100% feedback) and Mulloy et al. (25) (16/group with 70% feedback) show no benefit compared to no-feedback control groups, and Post et al. (28) (22/group) only shows a clear benefit in a transfer test. This pattern does not continue in the immediate effect studies, where feedback groups always outperformed no-feedback groups in at least one outcome measure. Otherwise, no clear pattern is visible regarding the time at which the tests were administered ("IT", "PT", "RT1", and "RT2" in **Table 2**) or regarding specific feedback regime parameters. While the studies in the immediate effect category yielded proportionally more positive results than the training effect studies, this was not statistically tested either and no risk of bias assessment was performed for this category, so this may be due to publication bias. The tendencies shown in the tests of the training effect category are further relativized by the concerns shown in the risk of bias assessment.

Because of the high risk of bias and because the included studies are too heterogenous in their design and especially their outcome measures, a statistical synthesis of the findings was not conducted. The risk of bias assessment revealed high concerns for all experiments in the training effect category except for Ericksen et al. (22) (some concerns) and Nekar et al. (27) (low concerns). Chan et al. (21) was considered to have high concerns with regard to feedback effectivity since the control group did not receive instructions to "run softer" in the intervention (effectively resulting in no intervention instead of a no-feedback intervention). All other high concern evaluations are already determined by domain 1 (underspecified randomization process) and domain 5 (no information due to lack of prespecified analysis plan). Any synthesis based on these results would therefore suffer from a very low strength of evidence. Attributing outcomes (positive or non-significant) to movement tasks, experience levels, or specific feedback parameter choices is not warranted, since any purported effect could be attributed to random chance or bias (induced by the specific selection or grouping criteria) rather than a generalizable property of motor learning.

Feedback regime recommendations from the literature

While **Table 2** may serve as a basis to find similar research to consider in future SAFT studies, the remainder of this section is devoted to summarizing recommendations made by the authors of included studies. These recommendations are not necessarily based on hard evidence, i.e., significant study results with a low risk of bias, and instead represent a collection of informed opinions to pay attention to in the future scientific investigation of SAFT.

Benjaminse et al. (20) concluded that the ideal feedback modality might depend on gender, with males in their study benefiting from visual feedback, whereas females instead might benefit from different feedback modes. Anson et al. (35) further mentioned that visual processing is slower and therefore more amenable to slow movements when compared to other modalities. Additionally, larger movements may be easier to detect with visual feedback than smaller movement details. Sigrist et al. (32) suggested that the effectiveness of concurrent feedback may not only depend on the complexity of the movement task, but also the complexity of understanding the task requirements. They stressed that different feedback modalities have different strengths, and further explain that concurrent visual feedback may be more suitable for instructing complex movement, whereas haptic feedback should be used instead for temporal guidance. Sigrist et al. (42) also discussed modality-dependent benefits (sonification for temporal aspects, visual feedback for spatial aspects). However, no significant benefit of multimodal over unimodal feedback was found in the study. They concluded that the selective advantages may be determined by the exact design of the feedback rather than being inherent to the modality itself.

Benjaminse et al. (20) also mentioned that providing subject views from multiple angles might improve the outcome, but that feedback with high complexity can be detrimental. Post et al. (28), however, explicated that the instruction to focus on the (previously defined) critical features of the movement task may be sufficient to avoid overwhelming the learner with the information presented in video (even without offering a video-specific interpretation). Rucci and Tomporowski (31) corroborated other results according to which video feedback without additional cues has little effect on skill acquisition. They emphasized that regardless of the feedback modalities used to deliver feedback, it should provide information on how movement errors can be detected (instead of only directing the learners' attention to the error). This complements Mononen et al. (24), who argued that it might be difficult to establish a link between the received feedback and the corrections that should be made. Teran-Yengle et al. (44) mentioned that real-time feedback can provide the learner with specific information that is not available with intrinsic feedback, thus encouraging exploration and discovery of alternative movement solutions.

Jones et al. (38) concluded that the practical effects of challenging correct feedback as opposed to threatening deceptive conditions should be further explored, and that their effects may ultimately depend on the performance of the learner as well. Washabaugh et al. (40) emphasized the importance of using external motivational tools, such as feedback, to increase both learning and training intensity when intrinsic motivation is lacking. Weakley et al. (41) stressed the importance of providing encouragement and feedback during resistance training, and further noted that the extent of the benefit and the most successful way of providing such encouragement may also depend on individual characteristics, particularly the degree of conscientiousness. In this line of argumentation, Rauter et al. (29) suggested that future studies should tailor feedback to the experience of the participants, that feedback should be changed over the intervention time to prevent studies from becoming monotonous, and, moreover, that such changes have the potential to reduce the induced feedback-dependency (Note that these recommendations specifically concern the planning of feedback in studies and may not be meant as a direct recommendation for feedback in practice). Also, Sigrist et al. (32) recommended to combine multiple modes of feedback and to use an intelligent feedback strategy that individually tailors feedback to preferences, learning rates, error patterns, feedback susceptibility, and performance.

Ericksen et al. (22) explicitly cautioned against using the proposed feedback without first examining retention and transfer effects. Post et al. (28) mentioned that their study could represent an example where transfer may be a more sensitive test of learning, and that self-selected scheduling of split-screen feedback facilitates motor learning under the right circumstances. Todorov et al. (33) explained that the goal of their study was to show that augmented feedback can give an advantage in a difficult multi-joint movement, so the characteristics of augmented feedback in their study were chosen with that goal in mind. They stressed that this consequently does not constitute proof that all the choices made were required to achieve a significant performance benefit. In other words, the chosen conditions were deemed sufficient, but possibly not necessary.

The other reports only mentioned intervention effects and general explanations, but did not state explicit, generalizable feedback regime or study recommendations based on their results.

Discussion

Summary and limitations

We aggregated information about the intervention and visual feedback regimes utilized in 26 studies on training complex, sports-related sensorimotor tasks. We additionally presented the authors' recommendations concerning feedback regimes. In general, studies were practice-oriented and therefore compared considerably different interventions with various feedback regimes, without making generalizability of results for specific feedback parameters a priority. Despite our efforts to increase homogeneity by applying restrictive inclusion criteria, this remaining heterogeneity and the differences between the measured outcomes make it difficult to relate effects of single parameters changes over multiple studies. For the studies with multiple main outcomes, taking one as the main outcome for such a comparison would be an arbitrary choice with a high risk of introducing bias. Consequently, a statistical synthesis of the effectiveness of different feedback parameters was considered inadequate. There were no clear indications as to which specific sensorimotor tasks or target populations might benefit from visual feedback, and where it should be avoided. Therefore, this review reported current trends regarding visual feedback regimes and their effectiveness in the research literature, but it could not provide strong evidence concerning specific feedback parameters. Moreover, when assessing the strength of evidence for or against the specific feedback design used, most included studies had either high concern according to ROB 2 or consisted of relatively small sample sizes per intervention group. As such, the described results should not be taken as definitive evidence, but rather as indications to take into consideration for guiding future research or practical implementation. For these reasons, we cannot give specific recommendations for practical SAFT system design and will instead summarize general considerations based on the designs and recommendations in the literature as well as giving theoretical guidelines to inform future research on SAFT system design.

By employing a strict search procedure specifically narrowed to sensor-based visual feedback, we set out to reduce the breadth of the study scopes *a priori*. These restrictive definitions were intended to facilitate objective evaluation but do not constitute a theoretical consensus. The exclusion of bimanual tasks, for example, was not based on research showing that these movements are necessarily simple tasks, but instead was a result of conservatively avoiding potential interference when including semi-complex tasks. Also, the boundaries between some other reported categories (e.g., concerning experience levels) should only be interpreted as rough indicators. Finally, the restriction to sensor-based feedback excluded functionally identical but non-sensor-based designs. For example, applying body-mounted laser pointers does not utilize sensors but provides the exact same information as a motion sensor and a display [cf. Stien et al. (47)]. On the other hand, raw video replay was included [e.g., Benjaminse et al. (20)] because of the camera sensor, which does not necessarily provide different information than a physical mirror [e.g., Roy et al. (48)].

While we believe we have covered the most important parameters in the design of visual feedback, there may be other important design variations in the remaining body of research beyond our search parameters and the three searched databases, especially in databases more related to sports. Based on the results shown here, we would not expect subsets with sufficient homogeneity to allow generalizable quantification of the benefits of specific feedback parameters even with a larger set of included studies. Including simple movement tasks, which tend to have more standardized testing and outcome measures, would not help with our main research question either because previous research has shown that the effects of feedback do not generalize to complex tasks (13–16). Be that as it may, our sample consisted of various settings in which visual feedback was used effectively, indicating that further usage and study of visual feedback seems warranted: In certain settings, visual feedback can have a positive impact, both on the immediate effects during training and on the learning and retention of complex sensorimotor tasks over longer periods of time.

Feedback regimes in the literature

We have seen a strong focus on knowledge of performance rather than knowledge of results. This may be explained by the fact that knowledge of results is often readily available (e.g., by looking at the point where a thrown ball has landed), so SAFT systems are not required in these cases. Moreover, designing concurrent knowledge of results may be more difficult and may not even make sense in non-continuous tasks. Indeed, the only case where we have seen concurrent knowledge of results was a cycling task where the result (total distance covered) is continuously updated. The benefits of KR or KP feedback have been discussed extensively in the literature, suggesting that it is a crucial aspect and that it should be considered when comparing one feedback intervention to another (49). However, there may be task goals and feedback regimes where the distinction is not so clear, particularly when execution of a prescribed movement without spatial error is the desired result [e.g., Koritnik et al. (39)].

Regarding the timing of feedback, we have seen little variation in feedback delay, with most feedback being simply described as concurrent or terminal. Sigrist et al. (17) concluded in their review that concurrent feedback is more beneficial as task complexity increases, so this could serve as a guiding principle. Anson et al. (35) argued that visual feedback is better for slow movements because visual processes take longer compared to proprioception. From this perspective, feedback delay is a spectrum rather than a binary property. This seems to be in contrast with the prevailing definition of concurrent or terminal feedback. We also note that in both concurrent and terminal feedback, delays in feedback could theoretically be added to encourage independent self-assessment and error prediction by the learner.

We found that feedback frequency was sometimes not reported, or at least not as a deliberate choice. As mentioned before, a reduced frequency could also be the result of tests during the intervention period. This, of course, should be taken into account when interpreting a feedback intervention from a study or using it in practice, as a different efficacy might be observed if the feedback training is not interspersed with non-feedback tests. In addition, strategies such as self-selected or error-based feedback could lead to an implicit, individualized fading mechanism, that promotes, for example, higher involvement and better transfer (50). If increased competence in the movement task through learning leads to fewer feedback requests or fewer errors exceeding the defined threshold, then this will effectively lead to less feedback received over time, as indicated by the vast discrepancies between average and maximum feedback frequencies in these regimes [e.g., in Post et al. (28)].

Feedback can be presented at different levels of abstraction and reliability. This may include, for example, ambiguities in representation, rounding of scores, combining multiple scores into one score, or over time (i.e., changing the resolution or specificity of the feedback). This can make it more difficult for the subject to interpret the results, introduce a threshold below which errors are imperceptible, or otherwise weaken the link between the measured quantity and the information conveyed to the subject. An example of deceptive feedback was given in Jones et al. (38), which is also a good example of using two different levels of abstraction: In addition to the more precise performance feedback provided by displaying distance traveled as a number, increased speed was also encoded in a complex graphical representation by moving an avatar faster through the environment. Taken in isolation, such complex feedback would not allow accurate differentiation of small changes in speed over time.

Finally, the most versatile parameter for visual feedback is the content of the graphical representation itself. We saw some complex graphics, but many of the included studies had relatively simple representations such as numbers, bars, and plots. The choice of visual feedback display format (such as plots, avatars, videos, etc.) seems to matter little. We would have expected much more variance in this area because it is becoming easier to develop such complex graphics and because commercial products with such graphics are ubiquitous, including exergames or virtual and augmented reality devices. This discrepancy could be explained by visual feedback becoming too complex for the learner to interpret effectively, or by potential confounding factors introduced with complex graphical representations that encode multiple variables simultaneously. Having said that, we have not seen any cases where the authors explicitly stated that the feedback was too hard to understand for the participants. None of the graphical representations were deemed too complex, and none of the quantities too abstract for the participants. As a result, we do not see a reason to restrict these parameters *a priori*. However, we should point out that the number of parameters conveyed at once were always rather small (i.e., at most three). It is not quite clear whether this was a purely scientific decision to control what the participants focus on, or whether this is a feedback design decision because participants may not be able to process or select from too much information at once. We would only expect the latter point to play a big role for concurrent feedback, since in the case of terminal feedback, there is ample time for the participant to study the information and select the most relevant parts in the terminal condition. A possible exception to the generally low number of parameters is present in video feedback: Depending on one's perspective, the scene can be interpreted as one parameter conveying the general silhouette or posture of the whole body, or it can be interpreted as containing a plethora of parameters including limb positions and joint angles. This might also explain the recommendations to guide the participants' focus with appropriate instructions, as this would affect the effective numbers of parameters to interpret.

We should also point out that the main goal of SAFT systems is to be beneficial for overall training, and comprehensibility of the provided feedback is only one aspect of this. It is unclear to what extent the feedback needs to be cognitively processed at all for it to help with the operationalization of certain movement parameters. After all, even if subjects find the visual feedback confusing or do not quite understand it, the feedback could in principle still have a positive effect because some (negative) patterns are still recognizable. This is more apparent in sonification, where understanding the parameterization may be more difficult than hearing when something about the movement is out of the ordinary. Another possible explanation for the relatively low diversity in the graphical content of the feedback are the rather uniform objectives of the feedback regimes we encountered: The feedback regimes were generally focused on direct error correction (with the error in question being directly related to the study outcome measures). Other possible objectives of feedback, such as guided exploration of the task-space through targeted variation of task and feedback parameters, remain largely uncharted. A more in-depth theoretical analysis of the movement tasks and training goals according to the four task-space learning mechanisms could encourage the examination of other feedback objectives.

Implications for the practical application of SAFT systems and future research

Implications for the application of SAFT systems in practice remain largely speculative. The main challenge to practically apply SAFT systems lies in identifying effective feedback regimes for specific sensorimotor tasks, and specific populations at specific stages of learning. The effectiveness of concurrent feedback may depend on the complexity of the movement as well as the complexity of understanding the task requirements. The optimal modality may depend on gender, speed of movement, and how large a movement is (i.e., visual discernability). There is some evidence that visual feedback is better suited for spatial task aspects (as opposed to temporal tasks), but Sigrist et al. (42) mentioned that this may be an artefact of simplicity of feedback design. In other words, designing intuitive feedback may be more straightforward if it has the same modality as the movement aspect, but that does not mean that otherwise a good design is impossible to find or that this feedback is inherently more effective. There may also be a tradeoff between feedback simplicity and the amount of information conveyed. Video feedback in particular may be too complex for the user, so additional, carefully formulated instruction is required. This guidance should ideally direct the user to correct the error and not just give information about the error, which necessitates a comprehensive understanding of the task and the involved control parameters. Finally, feedback can encourage the user to increase performance, but the effectiveness of this may be highly dependent on the user's preferences or skill level. The feedback should thus ideally be highly individualized and adaptive. When the motivational aspect is the main goal of the feedback, then the feedback regime might be regarded as successful even if it does not affect the overall training efficiency, as long as it does not hinder progress either.

In our opinion, the current research on feedback for complex skill learning does not support any sweeping statement for or against specific feedback regime parameters in practice. In this regard, not much has changed since the call for more intensive research on complex skill learning from Wulf and Shea (14) in 2002. It looks like visual feedback for complex movements at least does not lead to worse learning outcomes in most cases even if no explicit fading was implemented, provided that this is not due to publication bias. This lack of negative outcomes stands in contrast to feedback on simple movements [cf. the guidance hypothesis (11, 12)], which we interpret as corroborating Wulf and Shea's warning against using results from feedback studies with simple movement tasks to inform the feedback design for complex skills.

Whether visual feedback shows a significant positive effect or no significant effect at all seems to depend on the situation—how much this concerns the design of the feedback regime, the movement task, or the characteristics of the participant cannot be said with any certainty based on the current scientific literature. To better explain and predict the effectiveness of feedback in certain settings, standardized evidence is needed, so that a statistical meta-analysis that compares similar settings with low risk of bias becomes feasible. To this end, we call for future research to focus on obtaining clear definitions on what constitutes a complex coordination task and ideally finding task-category-dependent standardized coordination tests that can be utilized as main outcome parameters in different studies. After establishing a solid basis to build upon, systematic experiments varying only single parameters of the provided feedback for specific tasks would have the potential to produce prescriptive feedback design recommendations. Furthermore, generalizability of results from one outcome of interest to others in the context of augmented feedback training should be investigated: For example, it is not clear at the moment whether specific feedback design parameters, such as a reduced feedback frequency, would have the same effect in training for better endurance-running economy and training for increased weight-lifting performance. Interestingly, this need for more uniform, fundamental research on complex movement task learning with feedback mirrors the conclusion reached by Kal et al. (10) in a systematic review comparing the benefits of the implicit and explicit motor learning. This is a clear indication that this problem is not confined to feedback design studies, but rather points to a systematic issue with the design of trials investigating complex movement tasks in general, specifically the lack of trial and reporting guidelines as suggested by Kal et al. While there are useful reporting checklists for exercise studies, such as the Consensus on Exercise Reporting Template (CERT) (51), these checklists are not specific to feedback studies and only cover the reporting rather than the design of studies.

Theoretical considerations

In the absence of evidence-based guidance, we fall back on the theoretical background to inform future SAFT research to the best possible extent. First and foremost, it should be kept in mind that SAFT systems cannot be designed without considering the characteristics of the task and the instruction regime. Even if no explicit instructions are given to the learner, the way the feedback is presented during or after task execution potentially influences the learner's (implicit or explicit) task goals. As outlined in the introduction, SAFT system designers need to be aware of the subtleties of the well-established and researched motor learning approaches that lie between discovery learning and prescriptive, explicitly instructed learning. Only then can the designer leverage the real potential of systems to systematically assist motor learning during task space formation, exploration, differentiation, and (de-) composition. This is particularly important because instructions and feedback can cause shifts in attentional focus and influence learner motivation, triggering or hindering the learning of task specifics [e.g., compensatory effects (52)]. Unfortunately, the complexity of retrieving the correct instruction and feedback rises with the complexity of the task space. To tackle this issue, a structured approach to task understanding seems necessary. Naturally, domain specific knowledge, e.g., from experts in the field, in addition to evidence from similar previous research could provide a good basis for potentially fruitful feedback regimes. Complementary, functional task analysis (53) seems to be a well-suited approach to guide the identification of structure and functionally relevant features of the sensorimotor task without forcing the user to adopt a specific theoretical stance. Even if naturally the focus, functional assignments for specific modalities of the task's (sub-)actions are not limited to the biomechanical domain but can also be derived from anatomical, physiological, coordinative, perceptual, mental, or tactical perspectives on the sensorimotor task. As Hossner et al. (53) noted, these further functional justifications are based on the fact that a learner's perceptual-motor skills and psychological competencies shape individual task spaces. Hence, functional task analysis seems particularly suitable for the design of SAFT systems, as it automatically distinguishes (functionally irrelevant) style aspects from (functionally relevant) errors in the individual task solution. Both can be incorporated into the design of feedback—the latter as feedback that should be given to ensure correct and functional task solutions, the former as feedback that should be avoided to keep individual freedom and compensation potential high for the motor system and increase its robustness. Once the task space and relevant control variables are identified, the designer can begin to define the intended objectives of the feedback and instructions.

To define the intended objectives of the SAFT, a broad examination and prioritization of the potential benefits of feedback in the target setting is required. We describe some of these potential benefits for visual feedback here, but this list is by no means exhaustive. First, feedback can provide benefits simply by reducing monotony or making the learner more aware of their learning progress, which can, in turn, increase motivation (54). Second, feedback can be used to alter the goal-specifications or shift attentional focus (55). For example, adding an accuracy score in a throwing task might shift the learner's goal: Instead of trying to maximize the power output, the desired result might become movement precision or correct form, guiding the learner closer to an optimal solution. Such feedback may be necessary to guide the learner out of a local optimum in the task-space (4) or to encode variables related to injury risk in the optimization of a movement solution. Third, feedback could focus only on its immediate effect and not on lasting improvements. For example, correct posture and movement execution may be important factors for safety during strength and endurance training. In this case, it may even be beneficial to provide feedback to improve these parameters during each single training session, provided that the exerciser never has to perform these tasks without feedback, and they rather serve as basic building blocks for other skills. Fourth, visual feedback can be easily ignored by looking away, even if this is obviously not considered its primary intent. This may, however, be an advantage of visual feedback over other feedback modalities, as it allows for a form of self-selection that has been reported to increase the effectiveness of feedback and motivation (50). For an even more detailed discussion of the effectiveness of different types of feedback, we refer the reader to the pertinent review by Sigrist et al. (17). Since the intended objective of a feedback is critical for the design of the feedback regime, we additionally refer the reader to **Table 1** in Hossner and Zahno (5), where the specific roles of variance in different motor learning mechanisms are summarized.

There is not necessarily a fixed feedback regime that is optimal for all individuals. The optimal feedback strategy might even depend on the individual's daily mood, motivation, or physical condition, and it might change over a single training session with the level of fatigue. In addition, different aspects of the same task may be optimized in different ways, and tradeoffs could occur. For example, injury-prevention, speed, and jump height in volley spikes may be mutually contradicting goals that result in different optimal movement executions depending on the importance placed on each aspect.

Once a promising solution is found, a well-designed intervention study with fair controls is recommended to validate the effectiveness of the feedback intervention. If motivation is a primary objective of the feedback, even a null effect on learning rates may be considered a positive outcome, as it could mean that the motivational benefits can be reaped without impeding training progress. On the other hand, if the feedback-guided intervention is aimed at learning real-world skills in a training setting, transfer tests are needed to validate the effectiveness of the designed intervention, or at least, according to Teran-Yengle et al. (44), some sort of formal documentation of carry-over to normal life. When testing a novel training intervention with feedback, we strongly recommend three intervention groups: One with the novel training intervention with feedback, one with the novel training intervention but without feedback, and one as a classical control (no intervention or reference intervention). With such a design, the study can not only validate the effectiveness of the intervention, but it may also show the extent to which the outcome was influenced by the feedback provided.

Proposed strategy for SAFT system design in future research

Based on the literature reviewed and the theoretical considerations, we propose the following general strategy for designing SAFT systems in a scientific setting: First, clearly define the intended objectives of the SAFT. Second, conduct a functional task analysis to clearly identify functionally relevant control variables and error mechanisms. Third, determine options for initial feedback solutions based on prior research and domain-specific knowledge. Fourth, if needed to make an evidence-based decision, conduct small pilot studies to choose among different parameter options. Fifth, conduct a well-designed comparative study that includes transfer testing and a single clear main outcome measure. For novel training interventions with feedback, two control groups may be optimal: one with the training intervention without feedback, and one that does not receive the intervention. For established training interventions with novel feedback, a single control group getting the same intervention without feedback is sufficient. In both cases, we do not recommend designating a group receiving different feedback as the control group, unless the utilized feedback can be regarded as the gold standard in that setting. This procedure should support investigation of the potential benefits of a developed feedback intervention in practice as well as answering the question whether the feedback itself made a significant positive contribution to the overall outcome.

Conclusion

We compiled significant findings, utilized feedback regimes, and recommendations from a set of 26 studies on visual feedback in complex sensorimotor tasks with healthy adults. Although the current evidence base is insufficient to derive clear rules for or against the use of specific feedback regimes in complex sensorimotor tasks, the findings outlined in this review and the referenced research can serve as a basis for the initial steps in the process of developing a feedback regime for learning sports-related skills. Consideration of the properties of the sensorimotor task, the task instructions, the feedback regime, and the intended objectives of the feedback is critical. Because the evidence in the literature does not form a strong basis for an evidence-based feedback design guidance, the proposed strategy for future sensor-based augmented feedback training research is instead based on statements in the literature as well as the theoretical background. These considerations are only meant to inform feedback intervention studies in the interim. Standardized study design and reporting guidelines for motor learning research on complex movements, compiled by experts on motor control, are needed to direct future research in a way that will lead to a stronger scientific foundation that can adequately inform design decisions for sensor-based augmented feedback systems in practice.

Author contributions

HH and RK developed the research question. HH searched the databases. HH and JH reviewed the records in both phases and performed the data extraction. JH conducted the risk of bias assessment. HH drafted the first version of the manuscript, RK added the theoretical conceptualization. All authors provided critical feedback and corrections, contributing significantly to the research, analysis, and manuscript. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Publisher's note

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