Adaptation and evaluation of shoulder manipulator setup for assessing low extremity admittance and stiffness.

To explore the potential impact of fear of pain on posture control and athletic performance in the future.

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Abstract

Fear of pain is a psychological consequence that can hinder an athlete's ability to resume their previous level of performance or even return to their sport. A desire exists to examine the influence of fear of pain on both mechanical and neural factors involved in postural control, particularly focusing on the lower extremities. Previous literature shows associations between jump tests and fear of pain, but existing methods cannot differentiate between neural and mechanical factors, necessitating a different approach. Given the absence of a method to measure lower extremity reflexes and intrinsic properties similar to those observed in the shoulder, the original Proprio setup was adapted for leg perturbation to investigate its impact on lower extremity postural control. The adapted setup consists of a platform for participants to stand on, with coupling provided by a stiff ankle brace. A group of seven participants performed a slack trial and multiple stiff trials in both angled and neutral positions. Metrics such as admittance, stiffness, and coherence were calculated for each trial. The suitability of the setup was evaluated based on the selectivity of results across different trials and positions. Additionally, coherence was assessed, indicating the linearity which is necessary for future parameter identification. Notable differences in admittance and stiffness were observed between different legs and trials. This demonstrates the ability of the new setup around the Proprio to differentiate between various conditions. Comparisons with previous studies on shoulder dynamics showed similar findings, suggesting accurate measurement of leg dynamics. High coherence values indicated linearity, supporting the potential for future estimation of intrinsic parameters and reflex gains in leg dynamics. Considering these factors, the adapted Proprio setup appears suitable for analyzing lower extremity postural control. The next steps include expanding the participant group for significance and parameter identification, which will enable the investigation of the impact of fear of pain on intrinsic and reflexive properties using this setup.

Contents

1 Introduction

Sport and exercise generally have a positive effect on the public health and mental well-being of people. In the Netherlands, 53 per cent of the population of 4 years and older participate in sports, at least once a week. Of the adult population (18 until 64), 54 per cent exercise once a week or more often [1].

However, sports and exercise carry a risk of injury. In the Netherlands, data from 2022 reported that 11 per cent of individuals who engage in weekly exercise reported experiencing exercise-related injuries within the past three months [2]. On average, this group endured 2.4 injuries per 1000 hours of exercise. There are differences between genders, with men encountering injuries more frequently than women. Next to that, a significant proportion of injuries occur during popular activities such as football, fitness and running.

These injuries that occur through sports and exercise primarily manifest as physical trauma, which indicates damage to a component of the musculoskeletal system. Certain injuries can heal fast and without lasting consequences. On the other hand, in instances where an injury fails to heal adequately, it can profoundly impact an athlete's long-term quality of life, and in more severe cases, bring an end to a professional athletic career.

Beyond the physical trauma, injuries can also become more psychological. Research shows that a correlation exists between athletic injuries and negative psychological responses [3]. These responses encompass heightened tension, diminished self-esteem, and the onset of negative emotions such as depression and anxiety. The memories associated with the injury, including the pain and discomfort that was endured, often trigger a degree of fear or may evolve into a fear of pain or fear of re-injury [4].

Fear of pain is a psychological consequence that can hinder an athlete's ability to resume their previous level of performance or even return to their sport. This fear, also referred to as fear of movement or fear of re-injury is described as a specific fear of engaging in certain movements or activities that are incorrectly believed to be potential triggers for renewed pain or re-injury. In more extreme forms, fear of pain is referred to as kinesiophobia, which is characterised by excessive and irrational fear of movement. People suffering from kinesiophobia may not only avoid athletic activities but also everyday tasks and movements. This could lead to a sedentary lifestyle and associated health issues.

Moreover, in children, fear of pain can stem from growing pains, which is a common complaint. Growing pains are the most common cause of musculoskeletal pain in early childhood [5]. They may develop a fear of pain if they experience frequent or severe growing pains and may think that the pain can occur when exercising.

Therefore, the risk of injury is a concern that is not only important to different age groups and activity levels but could affect everyone in the population. Furthermore, fear of pain, whether it arises from injury, syndromes or growing pains, is a significant factor that can persist even after physical healing has occurred. This persistent fear may compromise an individual's willingness to engage in physical activities.

Fear of pain can arise as a response to a past injury, chronic pain or traumatic experiences. With athletes who have gone through injuries, they may become sensitive to pain and may overestimate the risk of re-injury. Research suggests that fear of pain is linked to decreased physical performance and lower levels of physical activity [6, 7]. The fear could result in avoidance behaviour for specific movements which could develop into compensatory movement patterns to avoid possible pain. With avoidance behaviour, the individual alters or restricts their movements to prevent discomfort. This can cause them to develop compensatory movement patterns, where they use other muscles or joints to avoid the affected area [8, 9, 10]. Over time, this can lead to disuse of the affected muscles or joints, weakening them further and making them more susceptible to injury [10]. Additionally, the compensatory patterns can place stress on other parts of the body, potentially leading to new injuries. This creates a cycle where pain leads to avoidance, avoidance leads to disuse and compensatory movements, and these could lead to pain or potential injury, continuing the problem (Figure 1).

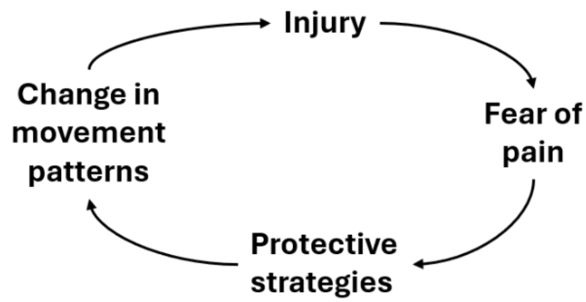

Figure 1: The cycle of pain avoidance and injury. An initial injury can lead to fear of pain, which in turn prompts protective strategies, such as avoidance behaviour. These protective strategies result in changes in movement patterns, potentially leading to further injury and continuing the cycle.

1.1 Objective

A desire exists to examine the influence of fear of pain on both mechanical and neural factors involved in postural control, particularly focusing on the lower extremities. The focus is on the lower extremities, since the literature is focused on the impact of fear of pain on athletic performance and that athletic performance is commonly measured with various jump tests. Previous literature shows associations between jump tests and fear of pain. However, existing methods in the literature cannot differentiate between neural and mechanical factors. This limitation necessitated a different approach.

The skeletal system, muscles, sensory organs and the central nervous system together form a closed-loop system capable of executing voluntary movements and responding to external disturbances [11]. The muscles function in this system as mechanical spring-dampers [11]. Reacting to external disturbances is crucial for maintaining posture and there are two mechanisms to counteract these disturbances. The first mechanism, the intrinsic properties, comprises a combination of elastic and viscous muscle properties resulting from muscle activation, along with passive elements, such as tendons and connective tissues [11, 12]. Consequently, intrinsic damping and stiffness depend on both passive tissues and muscle activation. The second mechanism involves the reflexive properties, which arise from muscle visco-elasticity due to sensory reflex loops [11]. These reflex loops provide sensory information; muscle spindles relay information about muscle stretch and velocity, while Golgi tendon organs provide sensory information about active muscle force [12]. Distinguishing between the intrinsic and reflexive mechanisms is challenging because muscle force comprises an intrinsic component stemming from muscle mechanics and a reflexive component, resulting from sensory feedback and muscle activation [12, 13].

System identification can be used to model the skeletal and central nervous system through interactions. Humans, in their everyday interactions, constantly engage with their environment [14]. These interactions are directed by predictions formed based on an inherent model that adapts to the environment through past experiences [14]. System identification can be utilised to understand and model these interactions. The human system is a dynamic system where the relationship between humans and the environment can be described with a feedback loop with inputs and outputs (Figure 2). With system identification, possible underlying models can be estimated from observed data [15]. This process allows for the prediction of the system properties and behaviours without the need for direct measurements of the parameters. By fitting mathematical models to observed input-output data, system identification provides a systematic approach to understanding and modelling complex systems [15].

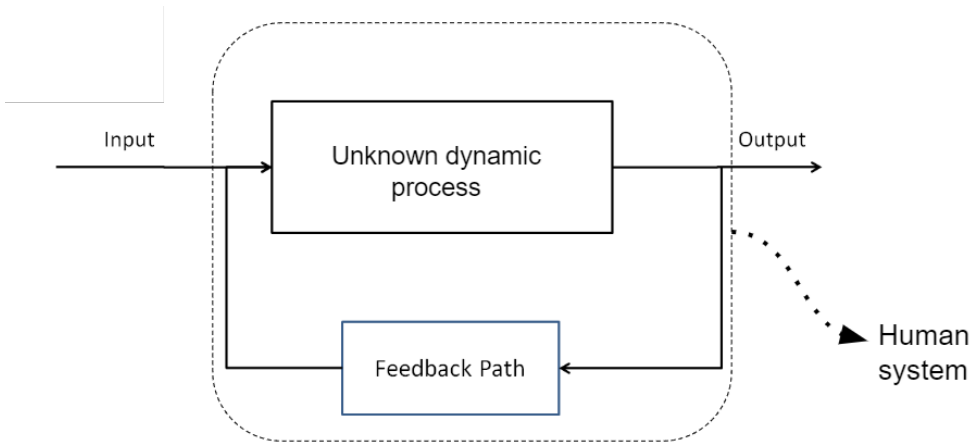

Figure 2: The human system described with a feedback loop with input and output. The feedback loop is possible due to sensory information.

Utilising the method of van der Helm, Vlugt and Schouten, it may be possible to investigate how fear of pain affects intrinsic properties, which involve both mechanical and neural factors, as well as reflexive contributions, which are primarily neural, during postural control with system identification. They utilised continuous random force inputs, applied by a manipulator called the Proprio. With system identification in the frequency domain, they identify the extent of reflex activity and intrinsic properties using various types or perturbations.

The intrinsic part of postural control can be estimated using wide-band (WB) disturbances, whereas the reflexive part could be estimated using narrow-band (NB) disturbances. With NB disturbances, the dynamics differ from those with WB disturbances due to the reflexive component. Co-contraction, which activates both antagonist and agonist, increases the intrinsic stiffness [12, 16]. Although this co-activation effectively counters all types of perturbations, it is energy-consuming [12]. With proprioceptive feedback, perturbations are detected by muscle spindles and Golgi tendon organs. With the sensory information, muscles generate a counteracting force via reflexes. While this method is energy-efficient, its effectiveness against high-frequency perturbations is limited due to time delay [16].

The Proprio is initially designed to perturb the human arm, and as a result, numerous experiments have been conducted focusing on the shoulder system [12, 17, 18, 19]. However, while extensive knowledge has been gained about the shoulder system using the Proprio, its application to the lower extremities remains unexplored. Given the absence of a method to measure lower extremity reflexes and intrinsic properties similar to those observed in the shoulder, the original setup around the Proprio was adapted for the use of perturbing the leg, to investigate the impact of fear of pain on lower extremity postural control. This leads to the objective of the study, which is to evaluate the suitability of a new Proprio-based setup for measuring lower extremity postural control.

For the design of the setup, several aspects must be considered to ensure efficacy, safety and adaptability. First, the setup should replicate the functionality of the existing setup, which is focused on the shoulder. The new setup should perturb the leg in one specific direction. Next, the setup mustn't restrict motion in the working direction. For the primary point of perturbation, a superficial bony structure has to be chosen. A bony structure provides a more stable and rigid foundation. When force is applied to bony landmarks, it can be more efficiently transmitted. Next, bony landmarks offer consistency across individuals, regardless of variations in soft tissue composition or body fat percentage. Furthermore, a rigid coupling between the perturbation point and the manipulator is essential. This guarantees the measurement of the intended system. Stability and support are eminent considerations in the design for participant safety and confidence during experiments. Robust construction and adjustable components provide a secure platform for participants, minimising the risk of injury and discomfort. Lastly, accommodating variations in participant height and stance is essential. Adjustable mechanisms, including height modification or adaptable support structures, enable participants of diverse anatomies to engage comfortably with the system, thereby enhancing the quality of the collected data.

1.2 Background

Within the literature, extensive research has been done on the biomechanical and neural adaptations resulting from fear of pain on sports performance [20]. The performance measures that have been frequently looked at are kinematic variables during movements, muscle activation patterns during jump tests, muscle strength and jump performance. Next to that, essential elements during movement are the range of motion and stiffness. Performance assessment has been primarily done with single-leg tests or drop vertical jumps, with the Tampa Scale of Kinesiophobia (TSK) measuring the fear of pain. The TSK serves as a quantitative measure for assessing fear related to movement, (re-)injury or pain.

Numerous studies have explored the relationship between fear of pain and the kinematics of the hip, knee and trunk during performance tests. However, the findings have been mixed, with some studies showing negative, while others have found positive correlations. The complexity arises from variations in study methods and the types of injuries considered, which make it challenging to compare findings and draw definitive conclusions [20]. Differences in tests include variations in the type of jump tests (one- or two-legged), and different single-leg tests (such as distance, crossover hop, triple hop, or timed jump). Additionally, variations exist in the height of the drop-jump test, where participants drop from a box and then perform a maximum vertical jump.

When considering muscle activation, significant positive associations between fear of pain and the activation of key muscle groups were found. In a two-legged landing scenario, the activation of the Gluteus Maximus during pre-activation was notably higher in individuals experiencing fear of pain. The Gluteus Maximus, responsible for external rotation and abduction of the femur, plays a role in mitigating the risk of anterior cruciate ligament injury [21]. The heightened pre-activation could potentially act as a protective mechanism, as internal rotation and adduction are recognised risk factors for anterior cruciate ligament injury. Studies indicate that preparatory muscle activation is particularly crucial in preventing knee valgus, a movement pattern associated with acute and chronic knee injuries [22]. Furthermore, during single-leg landing, decreased activation of the Vastus Lateralis was observed during landing and pre-activation phases [23]. This finding is consistent with kinematic changes where increased knee flexion was noted. Furthermore, there was increased activation of the Biceps Femoris during landing, although this was not observed during the pre-activation phase [24]. Moreover, individuals experiencing fear of pain exhibited higher co-contraction during landing, suggesting a compensatory mechanism. Co-contraction leads to increased knee joint stiffness, thereby enhancing stability. These findings shed light on the complex interplay between fear of pain, muscle activation patterns, and injury risk during dynamic movements [25].

Moreover, research indicates associations with the strength of key muscle groups, such as the quadriceps and hamstring, consequently influencing overall performance. While the amount of research was limited, studies have shown a decrease in lower extremity strength as fear of pain levels increases. One study measured the strength of the entire lower extremity rather than focusing on separate muscle groups. They found a consistent decrease in lower extremity strength with higher levels of fear of pain. This finding aligns with other studies that have observed decreased strength in individual muscles when fear of pain was present.

The observations may hint at protective strategies as a response to fear of pain, given that participants in these studies were no longer experiencing injuries. This indicates the interplay between psychological factors and physical performance.

2 Method

2.1 Instrumentation

2.1.1 Evaluating setup

To evaluate the new setup for the Proprio system for measuring lower extremity postural control, several requirements must be considered to quantify the setup's effectiveness and appropriateness. Firstly, the force disturbance signal must be within the range of the eigenfrequency of the leg. This is important for accurately measuring reflex gains during postural control. Reflexes are effective only for low-frequency inputs and are thus not active above the eigenfrequency [12]. Selectivity also needs to be taken into account. The setup should produce different results when participants perform different tasks, such as stiff versus slack postures or straight versus angled stances. This differentiation is important for measuring nuanced changes in postural control under various conditions. The frequency response functions (FRFs) of the admittance during different trials and tasks will be examined to validate the setup. Admittance represents the dynamic behaviour of the system in response to force perturbations. The following observations within these results will help evaluate the setup's suitability:

- An eigenfrequency peak should be observed in participants to confirm that the disturbance signal is within the correct frequency range. The eigenfrequency is the natural frequency at which a system oscillates when perturbed. At high frequencies beyond the eigenfrequency, reflexive feedback is limited [19].
- Observing a convergence of the NB trial results to WB trial results would suggest that intrinsic dynamics remain constant, whereas the reflexive dynamics change [12].
- Differences should be evident between the FRFs of different trials and between stiffness values of different positions. The observed differences would indicate the setup's sensitivity to variations in control tasks and positions.

It is important to note the small participant group (see Section 2.2), which prevents achieving statistical significance in the differences. Nevertheless, the qualitative and quantitative assessments will provide insights into the suitability and effectiveness of the new Proprio setup. In addition, the coherence will be assessed. The coherence gives an impression of the linearity of the system by showing how changes in one signal correspond to changes in another signal. The coherence is important for determining the setup's suitability for parameter identification using linear methods [12].

2.1.2 Manipulator with shoulder setup

The experiments in this study were performed with a linear hydraulic manipulator, the Proprio. This manipulator is described in Van der Helm et al. and is used for shoulder experiments [12].

The Proprio functions as a force-controlled actuator, which propels a piston to facilitate horizontal movement along the sagittal plane when the participant is seated facing the handle, as depicted in Figure 3.

The Proprio features a setup for the positioning of the participants, which is made for shoulder experiments. This setup comprises an adjustable seat that can be moved up or down, as well as a screen (Figure 3. The screen displays both the participant's hand position and a reference line set at zero. The objective for the participant is to maintain their position as closely as possible to this reference line. This strategy effectively prevents drift from the equilibrium position and positively influences the participant's effort and performance, resulting in smaller displacements [12]. Thus, the participants are performing a position task with force perturbations. Within this task, they generate muscle force to achieve a desired position. The muscle force is the combination of the intrinsic and reflexive pathways within the human control system [12]. Consequently, this muscle force generates a hand force, measured by a force transducer between the handle and the piston. The hand force counteracts the displacement induced by the external force disturbance [12].

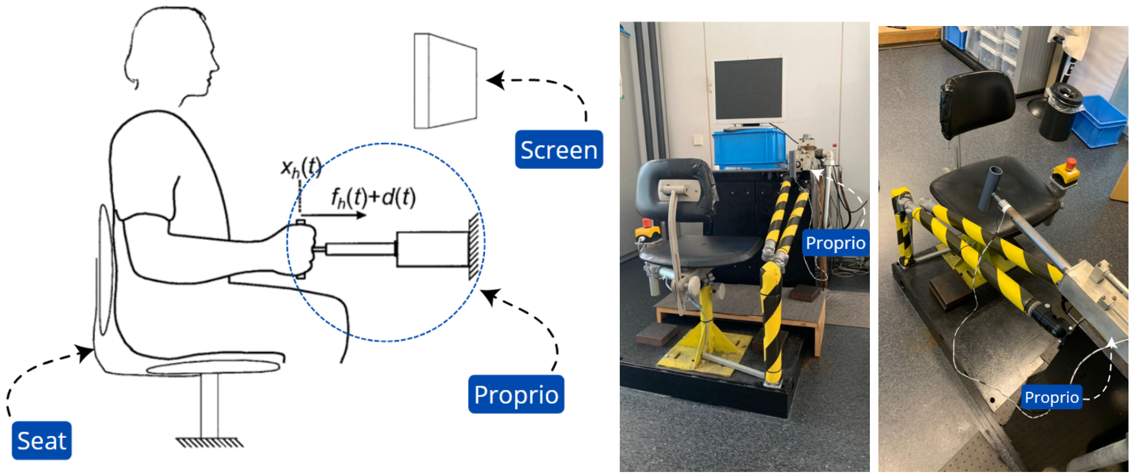

Figure 3: Original set-up around the Proprio when used for the shoulder. Illustration from Van der Helm, with hand force $f_h(t)$, disturbance force $d(t)$ and hand position $x_h(t)$ [12].

The stiffness and damping properties of the Proprio can be adjusted to adapt to different environmental conditions. However, since this study is focused on the suitability of the Proprio with the leg, the environmental variables remained constant. Throughout the experiment, both stiffness and damping parameters are fixed at zero for each trial, thus maintaining a consistent experimental condition.

Figure 4: The experimental set-up shown in the lab in front of the Proprio. It is not positioned correctly here, due to the presence of the seat from the old set-up, obstructing its proper placement.

2.1.3 Leg setup

As depicted in Figure 3, the setup around the Proprio is made to allow perturbations to the arm and shoulder joints. To use it on the leg and perturb the hip joint, a new setup has been built around the Proprio so that the ankle comes to the height of the piston of the Proprio. This new set-up has been pictured in Figure 4.

Considerations The set-up had to replicate the functionality of the existing set-up used for the shoulder. Several considerations are involved in achieving this. Firstly, the force perturbations should be directed appropriately. In the case of the shoulder, the piston moved in the sagittal plane of the person. For the current purpose, the focus is on the abduction and adduction of the hip. This movement occurs in the frontal plane, which means that the participants will be positioned sideways with regard to the device rather than facing it. The decision to focus on measuring the leg was made due to the existing literature's emphasis on jump testing when measuring sports performance. Perturbing in the frontal plane allows for isolating the hip movement and keeping the knee and ankle stiff.

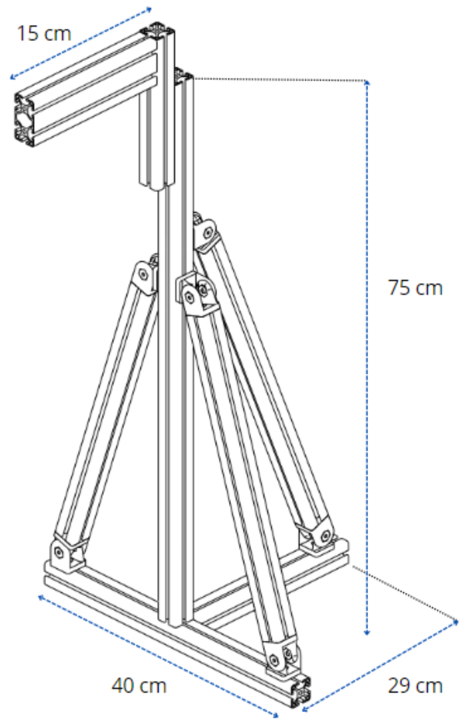

Figure 5: Sketch of the seat, made out of Item profiles.

The force perturbations are applied at the ankle, which features a prominent bony landmark easily accessible beneath the skin's surface. Participants are positioned on a platform designed to elevate the person, so the ankle comes to the level of the piston. To ensure stability and support, both a seat and a railing are provided on the platform. The seat, as depicted in Figure 5, is constructed using Item profiles. It consists of multiple beams, with three of those beams providing support to the sides and back. This ensures stability. For comfort, a bike saddle is placed on top of the seat. Recognising the variation in participant's heights and stances across the trials, the seat is adjustable in two directions on the platform: up and down and sideways. This adaptability accommodates individual differences, ensuring optimal positioning for each participant during the interaction.

Additionally, the setup should not restrict motion in the direction of the perturbation. In the setup, the feet are positioned side by side. Subsequently, the perturbation signals must be generated within a range where position deviations are constrained enough to prevent any contact between the feet.

Finally, there must be a rigid coupling between the ankle and the manipulator to ensure that the correct system is being measured. Facilitating the interface between the participant and the manipulator, an ankle brace with a rigid base connects the participant's ankle to the piston. Since movement of the brace against the ankle was still possible when only using the ankle brace, an additional band of stiff Velcro was introduced. This extra piece of Velcro was secured around the ankle beneath the attachment to the Proprio, which minimised potential movement between the brace and the ankle, enhancing control during the experiment.

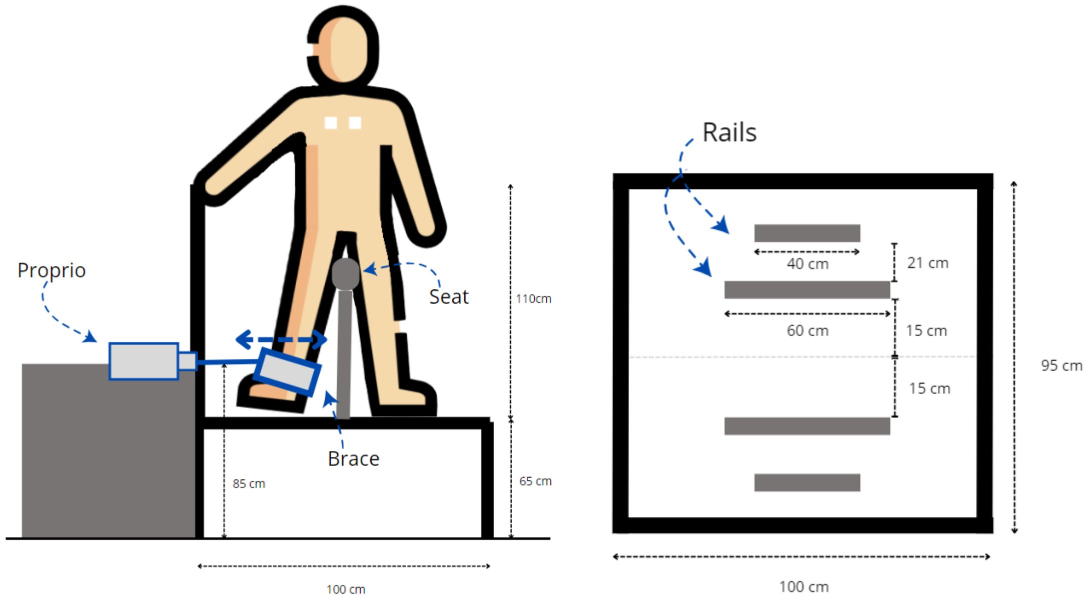

Figure 6: The experimental set-up, showing the participant next to the manipulator. At the end of the piston, an ankle brace is attached. The participant interacts via this ankle brace with the manipulator. The chosen dimensions are shown. The right image shows the top-down view of the platform with the rails of the seat for sideways displacement. The image on the rights shows a top-down view of the platform, illustrating the rails used for laterally displacing the seat.

Dimensions The dimensions for the leg setup were chosen based on standardised proportions of the male relative to its height and available space around the Proprio [26]. Considering that the average height of a man in the Netherlands is approximately 185 cm, this value was used to define the dimensions [27]. All chosen dimensions are pictured in Figure 6. All selected dimensions are illustrated in Figure 5, 6 and 7.

The piston of the Proprio is positioned at a height of 85 cm from the floor. The platform is situated 20 cm below that, resulting in a platform height of 65 cm. This configuration accommodates the brace and ensures ample space for the foot to move above the floor. The platform dimensions are set at 100 by 95 cm, which is bigger than the existing platform. This decision was based on the available space around the existing platform, ensuring that individuals can walk around it comfortably to use the computer of the Proprio. Additionally, the platform size allows individuals to stand with their feet wide apart.

The railing that provides stability is positioned at a height of 110 cm above the platform. This height is consistent with the standard height of a ballet barre. This is slightly higher than the hip height for an average height man of 185 cm (hip height = body height * 0.53 = 0.98 m), providing more stability compared to a lower bar.

Regarding the seat height, it was determined that the seat should accommodate individuals with heights ranging from 170 to 190 cm. This corresponds to a hip height ranging from 90 to 100 cm. With a saddle height of approximately 15 cm, the current seat can be adjusted to provide a seating height ranging from a minimum of 85 cm to a maximum of 103 cm. The seat can be laterally displaced, as depicted in Figure 7, to accommodate the two different positions; angled and straight. The sideways displacement of 17 cm is primarily determined by construction limitations rather than theoretical considerations. The rails on which the seat is placed do not allow greater displacement. Initially, the sideways displacement was designed to provide flexibility in adjusting the seating positions to suit different users, rather than to introduce a large angle α. With leg lengths of individuals ranging from 90 to 100 cm, the displacement to the angled position corresponds to angles of approximately 10 degrees.

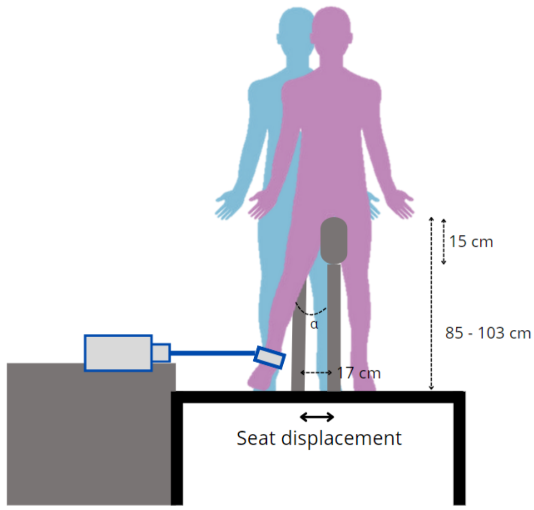

Figure 7: Sideways displacement of the seat. The seat can be displaced sideways by 17 cm to transition between the neutral (blue figure) and angled (purple figure) positions. The working leg, which is the left leg in the figure, remains in the same position. The sideways displacement of the seat creates the hip angle α.

2.2 Participants

This study involved five young, healthy men with a mean (SD) age of 23.6 (2.2) years. Additionally, two middle-aged men, with a mean (SD) age of 43.5 (4.9) years, also took part in the study. Before their participation, all participants gave informed consent. Within the older age group, participants reported having experienced sports injuries in the past. Despite these previous injuries, they are now engaging in regular exercise. However, as a consequence of their past injuries, they approach exercise more carefully.

2.3 Procedure

The experiment consists of 45 trials, lasting 30 seconds each. Before performing the trials, participants perform a few test trials with different disturbance signals than those used in the experiment to become familiar with the task and equipment. During the experiments, participants followed two different tasks.

(A) Minimise displacement of the leg. Participants are instructed to maintain their position, as displayed on the screen. To prevent drift, they should stay as close as possible to a designated line also presented on the screen.

(B) Relax the leg. Participants are instructed to minimise effort and relax the working leg, thus refraining from reacting to any disturbances applied by the manipulator. At the onset of each trial, participants adjust their position to align with the line on the screen, ensuring a consistent starting point for each trial, before relaxing the working leg.

2.3.1 Positioning

In this new set-up for the Proprio, two positions are used to elicit different responses from participants.

- Neutral position. In this position, the working leg is allowed to hang straight down, mimicking the original set-up of the Proprio where the upper arm is positioned vertically. This alignment facilitates a baseline assessment.

- Angled position. The seat is shifted 17 centimetres to the side, in the direction away from the piston, which induces an angle at the hip joint in the frontal plane. This position comes from the initial research question to investigate the effects of fear of pain during postural control. This positioning choice originates from the initial aim of the study, which was to investigate the effects of fear of pain.

To evoke this fear of pain, participants needed to be placed in a position that could potentially trigger this fear without causing physical discomfort. Selecting this specific angle was based on the assumption that participants might experience fear associated with movements involving their hip region in the frontal plane. This position was maintained to examine differences in measurements and participants' experiences with different positions.

During both positions, participants are instructed to actively extend the measured leg to maintain a straightened knee and a stiff leg.

2.3.2 Disturbance signals

The disturbance signals used are continuous and random. This is to prevent anticipation of the human. When a deterministic signal, which can be represented by a mathematical expression and be predicted at any time instant, is used, a human can predict and counteract the perturbation [16]. When the disturbance signal is unpredictable, the proprioceptive feedback properties of the human control system will be used [16]. There will be 2 types of disturbance signals used (Figure 8) [12].

- Wide bandwidth (WB) signal. This signal comprises frequencies ranging between 0.5 and 19.8 Hz.
- Narrow bandwidth (NB) signal.
  - Type 1. For the type 1 NB signals, the lowest frequency is consistently 0.5 Hz, while the highest frequency varies between 1.2 and 3.7 Hz. This results in six type 1 NB force disturbances.
  - Type 2. For the type 2 NB signals, the bandwidth of the signal is always 0.3 Hz. The centre frequency varies between 1.3 and 7.1 Hz. This results in eight type 2 force disturbances

All disturbances are applied three times during task A, resulting in 45 trials. Next to that, the WB disturbances were also applied with task B for 3 trials.

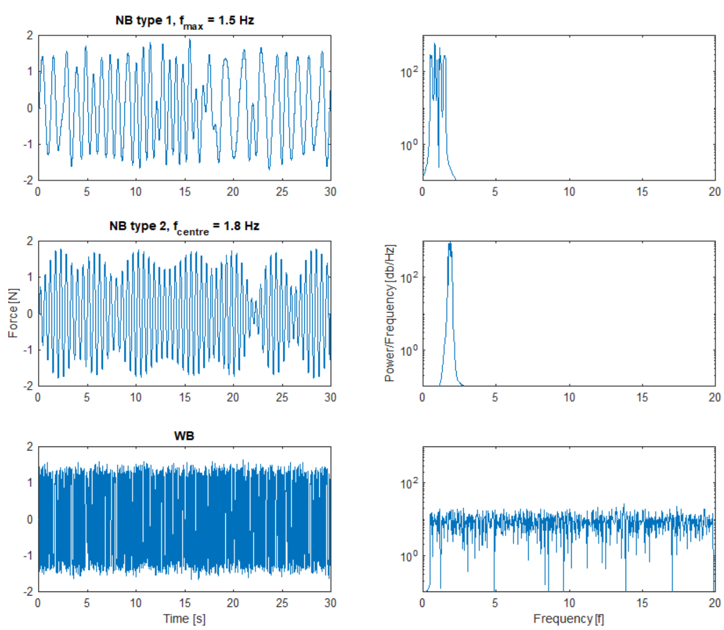

Figure 8: One example of each force disturbance signal type utilised on the left. Upper: NB signal type 1 (0.5 - 1.5 Hz). Middle: NB type 2 signal (1.65 - 1.95 Hz). Lower: WB signal (0.5 - 19.8 Hz). The right figures show the power spectral densities of the left signals.

2.4 Data processing

2.4.1 Signal recording

During the experiment, the following signals were recorded: ankle position $x(t)$, ankle force $f(t)$ and the external force disturbance $w(t)$. Each trial lasted 30 seconds. The signals were recorded with a sampling frequency of 2500 Hz. During data analysis, the first four seconds of each recorded signal were discarded to eliminate any transient response. Transient response refers to the initial reaction of the dynamic system to the sudden disturbance.

2.4.2 Non parametric analysis

For the non-parametric analysis, the frequency response function (FRF) for each trial was estimated following the method of Van der Helm et al. [12].

First, the recorded signals were averaged in the time domain. These signals are then transformed to the frequency domain, using the Fourier transform. Then, the power spectral densities (PSD) are calculated with the following formulas:

$$S_{wf}(f) = W(-f)F(f) = W^*(f)F(f) \tag{1}$$

$$S_{wx}(f) = W(-f)X(f) = W^*(f)X(f) \tag{2}$$

The PSD quantifies the spectral content of a signal, showing how the signal's power is distributed across various frequencies [28]. To improve the estimate and reduce variance, the spectral densities are averaged. This is done over four frequency bands.

Next, the following FRF was calculated, using the PSD:

$$H(f) = -\frac{S_{wx}(f)}{S_{wf}(f)} \tag{3}$$

The $H(f)$ is called the admittance. The admittance measures the system's responsiveness to force perturbations. It describes the limb's response following a perturbation, indicating the position deviations that result from a given force [16]. When admittance is high, the system is more compliant, allowing greater position deviations in response to applied force compared to a stiff system with low admittance. [16]. The computation of the admittance aims to detect differences in behaviour, enabling the identification of differences in system dynamics.

For the equation, it is assumed that the noise present in the system is uncorrelated with the force disturbance $w(f)$. The admittance is calculated per participant, per leg, and per trial.

Leg stiffness is a measure of resistance given and incorporates information about both the motion and the force encountered by the body [29]. Calculating the stiffness facilitates comparisons with findings from other research and methods. Leg stiffness can be calculated by dividing the applied force by the change in position [29]:

$$k = \frac{F}{\Delta x} \tag{4}$$

For which the F is the applied force of the Proprio and the Δx is the change of position of the leg, measured at the ankle.

Alternatively, stiffness can also be calculated using the impedance. The impedance is the inverse of the admittance and describes a change in force due to a position disturbance [16]. When analysing a Bode plot of a standard second-order system, a first segment can be drawn at lower frequencies up to the eigenfrequency. The eigenfrequency is characterised by a peak in the Bode plot and represents specific frequencies at which a system is prone to vibrate [30]. This initial segment, as seen in Figure 9, is primarily influenced by stiffness, allowing for the direct derivation of leg stiffness from the impedance Bode plot [31].

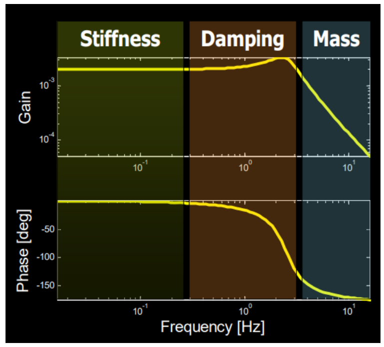

Figure 9: Bode plot of a standard mass-damper-spring system [31]

The impedance can be estimated using the formula [12]:

$$H_{impedance} = \frac{S_{wf}}{S_{wx}} \tag{5}$$

Subsequently, the mean value is calculated from the lower frequency segment of the graph, preceding the eigenfrequency. The cut-off frequency for the segment for which the stiffness will be calculated will be the same for each participant and trial and this cut-off frequency will be determined afterwards.

2.4.3 Measure of reliability

As a measure of reliability, the coherence is used. The coherence is defined as:

$$C_{xy}(f) = \frac{|S_{xy}|^2}{S_{xx}(f)S_{yy}(f)} \tag{6}$$

Coherence provides an impression of measurement quality, reflecting the linearity of the measurement. It ranges between 0 and 1, where higher coherence suggests more linear system behaviour. A coherence value less than 1 indicates the presence of noise, non-linearities, or time-varying behaviour in the system [14]. The coherence is considered low when below 0.8 [12].

3 Results

3.1 Performance leg setup

After some test runs with the new setup, a few adaptations were made to ensure optimal measurements and experiments. Firstly, the platform had to be raised slightly as the ankle was too low for the piston to comfortably accommodate the participants. When the ankle was too low, the hips had to be tilted to get in line with the piston, which was uncomfortable. Heavy blocks were placed underneath the legs to raise the platform by approximately 10 cm. This adjustment brought the distance between the piston and platform close to the distance between the ankle and the floor, which is around 10 cm for a man with a height of 185 cm. Furthermore, the measurements were initially not corrected for position deviations, which were small at the beginning. The goal was to achieve similar angles to those when using the Proprio with the arm. Assuming an average arm length of 35 cm and a position deviation of around 1 cm, this corresponds to an angle of around 1.6 degrees with the upper arm. To achieve a similar angle for the hip, a position deviation of around 2.5 cm was required. To achieve this, the gain for the force disturbances, initially set at 3, was adjusted. The gain was doubled for some of the NB type 2 signals and almost doubled for the WB signal. The gain for other signals was increased by 1 or 2. This did not result in significantly greater position deviations. Despite this, the higher gains were used for further measurements.

3.2 Participant characteristics

The main characteristics of the participant group are seen in Table 1. The participant group is rather active and participates in a diverse range of sports, like badminton, hockey, running and lifting. Participants 1, 2, 3 and 4 performed the neutral task with both the right and left leg, while participants 5, 6 and 7 performed the neutral task only with their dominant leg. Participants 1, 3, 5, 6 and 7 performed the angled task.

Age	Height	Activity level		Dominant side
		Days	Hours per day	
29 ± 10 years	184 ± 4 cm	4.9 ± 1.3	2 ± 0.6	7 right, 1 left

Table 1: Main characteristics of the participants in this study.

3.3 Spectral analysis

The power spectrum in Figure 8 reveals that the WB disturbance signal has power across all frequencies. This coverage ensures that the observed decrease in the admittance function is attributable solely to the characteristics of the system that is being tested, rather than limitations in the input signal.

Similar behaviour is observed for all participants for different conditions. Figure 10 shows the admittance function, the phase and the coherence of two participants for the WB and NB conditions in the neutral position as examples. Among the participants, some exhibited a distinct peak around the eigenfrequency in the gain graph, while others exhibited a smoother curve with a broad peak. In all cases, the phase consistently descended towards -180 degrees around the eigenfrequency before increasing again to -45 degrees. Furthermore, the coherence values remained consistently high for all participants for the NB trials, not going below 0.95. In the case of the WB trails, there is a higher degree of variability observed in coherence values. Despite this increased variability, coherence values remain consistently greater than 0.8.

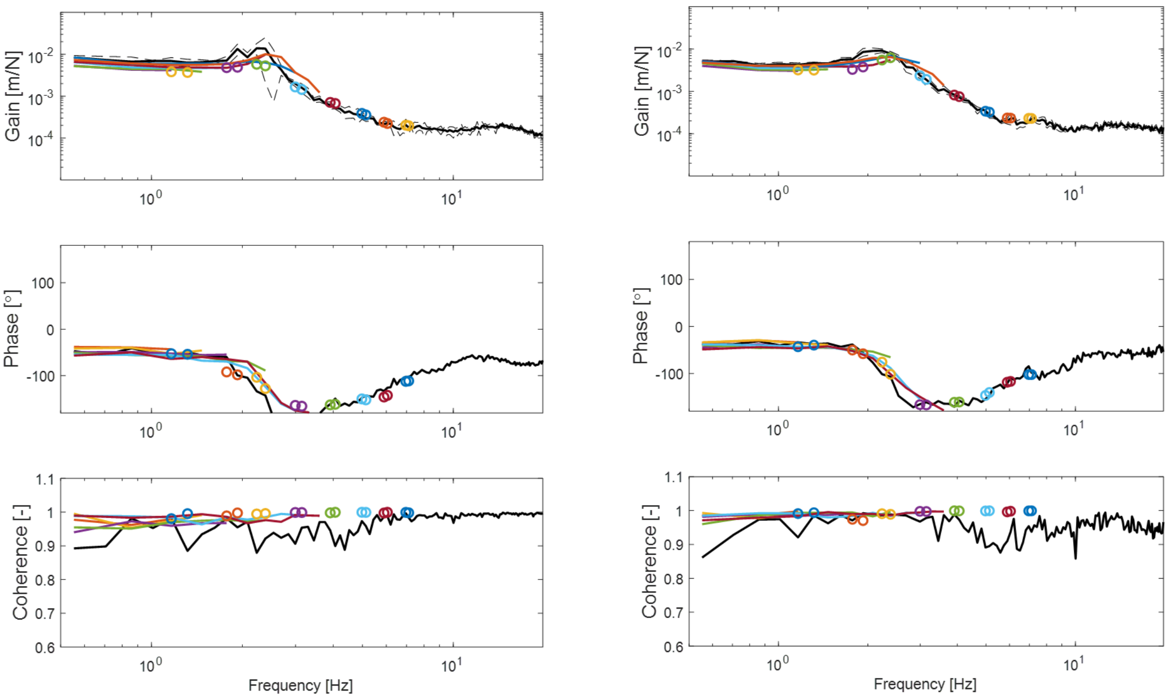

Figure 10: Frequency response (admittance function), phase and coherence functions of participant 1 on the left and participant 3 on the right, both of the right leg in the neutral position. Blue solid lines are from the WB condition with the dashed lines the standard deviation. Other coloured lines are from the different NB conditions, plotted without the standard deviation.

Figure 11 depicts the admittance functions of both the stiff trial, which was conducted in two positions, and the slack trial. Across all participants, a consistent trend is observed: the sequence of admittance functions remains unchanged. Specifically, the slack condition consistently displays the highest admittance, followed by the neutral trial, while the angled trial consistently shows the lowest admittance. This consistent pattern persists among all participants who have completed both the neutral and angled conditions.

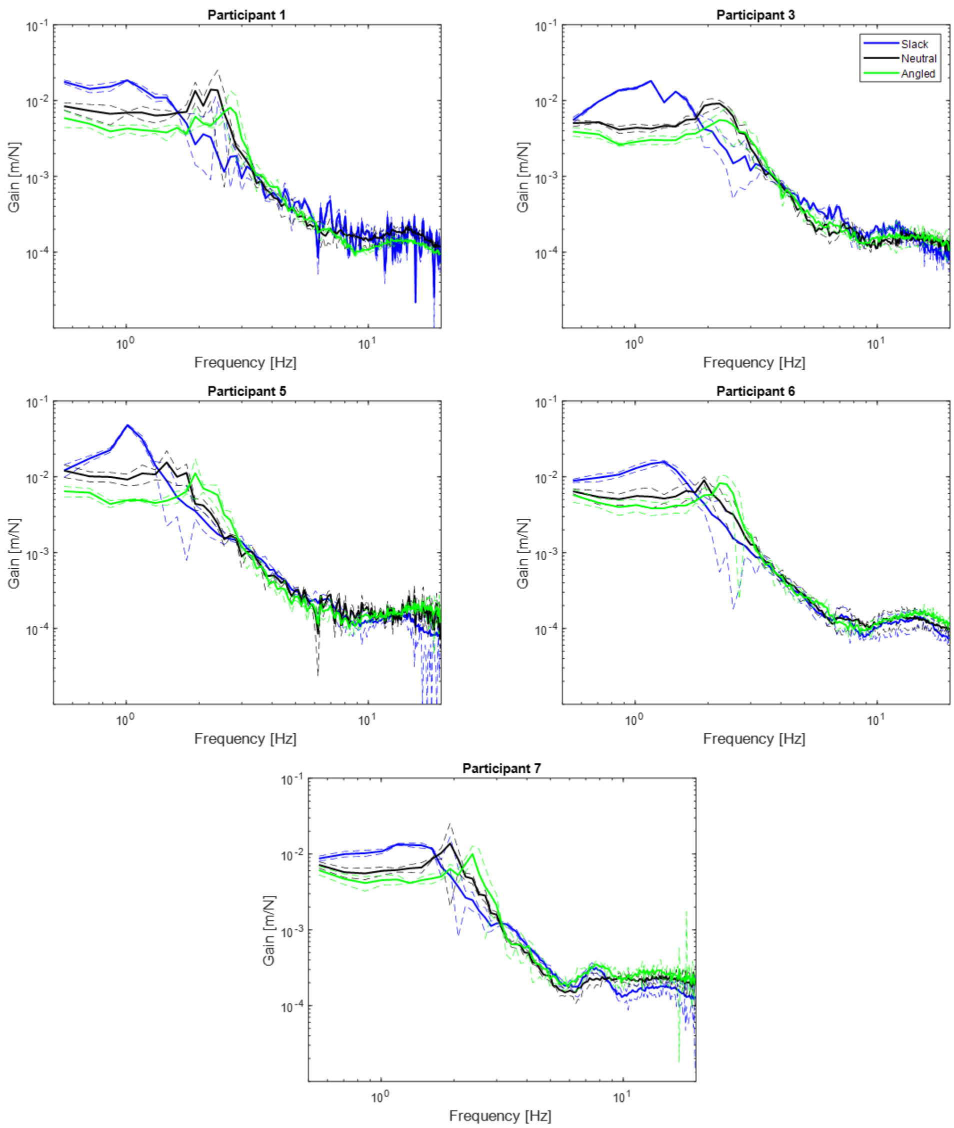

Figure 11: Admittance functions (solid lines) of the trials in slack, neutral and angled conditions with their standard deviation (dashed lines).

3.3.1 NB versus WB

Figure 12 shows the admittance functions for both NB and WB trials conducted by participants using their left and right legs. Each NB trial is depicted separately in the graphs. Notably, for the NB type 1 trials, lines are drawn. For the NB type 2 trials, which are characterised by a very narrow bandwidth, the functions are drawn as points in the graph. Across all participants, the NB FRFs exhibit a trajectory that follows that of the WB trial. However, a notable distinction is the absence of the peak around the eigenfrequency within the NB trials, which is present in the WB trial functions. In general, the functions of the NB trials tend to lie slightly lower than those of the WB trials for most participants, although they typically remain within the standard deviation of the WB function. An exception occurs with Participant 5 in the neutral position with their dominant leg, where the NB function falls below the standard deviation of the WB function.

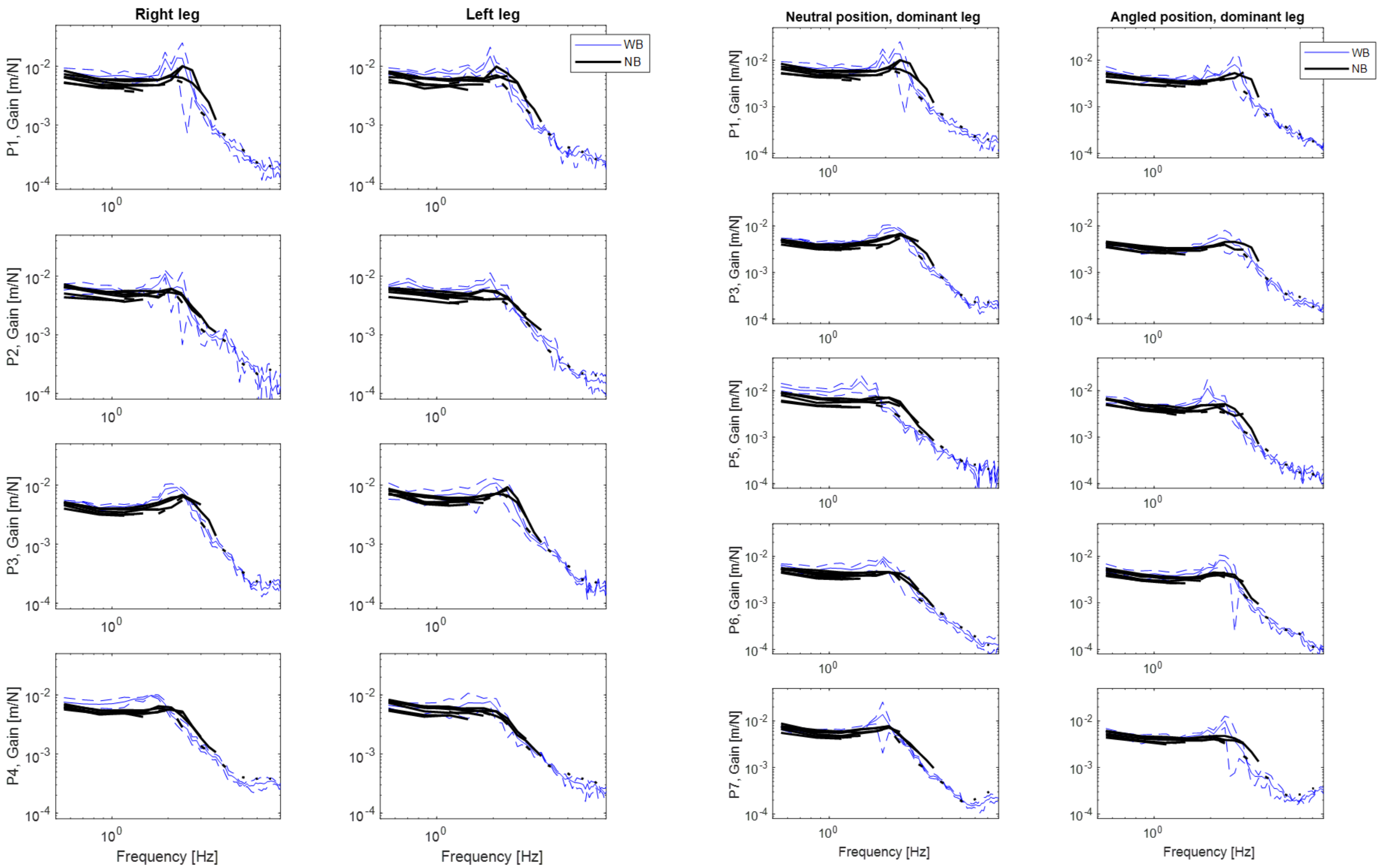

Figure 12: Admittance FRFs from both NB and WB trials are presented. The two leftmost columns correspond to the right and left legs of participants 1-4. while the two rightmost columns display the FRFs from the neutral and angled position of the dominant leg for participants 1,3,5,6 and 7. The WB functions are depicted by blue lines, with dashed lines indicating the standard deviation. Conversely, the different NB conditions are represented by black lines.

3.3.2 Leg stiffness

Stiffness values across the various conditions are provided in Table 2. The stiffness values were calculated using a cut-off frequency of 1.5 Hz. For all participants, the eigenfrequency peak occurred after this frequency, ensuring that stiffness values were accurately calculated. When analysing the results, no trend can be observed in the difference in stiffness between the left and the right sides among participants 1 to 4. However, participants 1 to 3 exhibit slightly higher stiffness on their right side in comparison with the left. Nevertheless, these differences appear marginal and are likely non-significant, considering they fall within the standard deviation of the stiffness values, except for Participant 3. Conversely, participant 4 demonstrates a higher stiffness on the left side compared to the right. Furthermore, when comparing the differences between the neutral and angled positions, every participant consistently shows greater stiffness in the angled position.

Participant	Stiffness			Difference	
	Right	Left	Angled	Right - Left	Neutral - angled
1	143 ± 13	126 ± 12	231 ± 34	17	88
2	172 ± 12	161 ± 17		11	
3	219 ± 18	142 ± 11	323 ± 43	77	104
4	132 ± 11	156 ± 10		-24	
5		91 ± 14	197 ± 30		60
6	182 ± 14		234 ± 30		52
7	161 ± 14		217 ± 26		56

Table 2: Stiffness value for every participant and their performed conditions. Differences between stiffness values are calculated for the difference between the right and the left leg, both in neutral position, and the neutral and angled position, where the dominant leg stiffness has been used for neutral stiffness value.

4 Discussion

Fear of pain is known to influence performance in physical activities such as sports and rehabilitation. However, understanding its impact on reflexive components and intrinsic properties remains challenging due to measurement limitations. To address this gap, the suitability of a new setup centred around the Proprio needs evaluation.

4.1 Performance leg setup

The seat exhibited a slight sway, despite the three support beams intended to provide stability. In future iterations, modifications to the seat design should be considered to address this issue. Another observation was that the platform also exhibited some small degree of sway. This sway could have influenced the measurements, so it is important to minimise it for more accurate results. This could be by strengthening the platform legs, for instance, by implementing x-bracing. This is a commonly used technique in construction to reduce sway and increase stability, which could effectively mitigate this issue [32].

Additionally, enlarging the platform to accommodate a wider range of leg angles could be beneficial. In this study, the angle was relatively small. With the 17 cm displacement of the seat, the angle was approximately 10 degrees in the starting position. Despite this small angle, participants reported discernible differences in their level of control when being in this angled position. Furthermore, since this study did not involve participants with a fear of pain, the largest hip angle possible construction-wise was chosen. However, for future studies, it may be beneficial to explore different angles to assess their impact on the measurements. Expanding the platform would allow for testing at various leg angles, providing a more comprehensive understanding of how different leg angles affect participants' perceptions of control and stability during the experimental tasks, enhancing the overall effectiveness of the set-up. To accurately determine the angle of the leg, a goniometer can be employed. Goniometers are utilised to measure the range of movement of joints by determining the maximum angle a joint can achieve. In the present study, the angle is estimated using seat displacement and an approximation of leg length. However, using a goniometer would provide a more precise measurement of the starting angle.

Furthermore, the gain increase mentioned in Section 3.1 did not result in the desired position deviation. Although the increased gain led to some increase in position deviations, these increases were mostly observed with the low-frequency disturbances. At the high frequencies, the position deviation remained small (less than 1 cm). When there are only small position deviations, there is a chance that the movements stay within the short-range stiffness. The muscle short-range stiffness is a property of the muscles that causes a rapid increase in muscle force due to stretch [33]. This likely has an effect on the dynamics in the initial response to the force perturbations [33]. The short-range stiffness can provide stability during the initial response against small, external perturbations [33]. With the small position deviations, it is possible that the muscle was never extended beyond its short-range stiffness. With larger movements, the muscle gets stretched and becomes more compliant [34]. Hence, in interpreting the results, it is important to consider the possibility that the muscle was operating within its short-range stiffness. With larger position deviations, the leg might exhibit reduced stiffness, leading to increased admittance. For future research, the gains could be adjusted manually before each trial, as was done by van der Helm et al. [12]. They modulated the power for each trial to maintain constant position deviations.

4.2 Spectral analysis

The mechanical admittance describes the displacement x in response to the input disturbance force w. The FRFs observed in this study resemble those of standard second-order systems. The lower frequency range is primarily dominated by stiffness, whereas at higher frequencies, resistance increases due to inertial properties such as mass, resulting in a decrease in admittance [12]. Across all participants, a peak is evident in the FRFs of the WB trials, indicating an underdamped system. Upon examining the phase, a consistent phase lag is observed, as the phase consistently remains below zero. This implies that the output signal x lags behind the input signal w, suggesting that participants consistently reacted late to the disturbance force. This delay could stem from internal time delays and the inability to predict force disturbance.

The part after 10 Hz has a high coherency, which could be explained by the coupling between the ankle and piston. At those high frequencies, a lighter system absorbs the movements caused by the disturbance force, which dominates then the FRF. Within the coupling between the ankle and piston, some small movements were still possible, which could make up this lighter system.

High coherence values indicate the absence of noise and suggest that the system behaves linearly [14]. In such cases, system identification can be effectively performed. While the coherence values are consistently high across the different trials, interpreting coherence here requires caution. Considering that the perturbation signals are multi-sine, this may lead to overestimation of coherence, particularly when averaging is based on a limited number of frequency bands. Taking into account that we have already averaged over a limited amount of bands, 4, coherence is prone to overestimation. Furthermore, coherence tends to be overestimated with a poor number of samples included in the calculation [35]. In the case of the NB trials, especially the type 2 trials, coherence is exceptionally high. However, the type 2 signals contain a limited amount of data points after averaging for calculating the FRF, which further contributes to this tendency for overestimation.

The differences between the neutral and angled conditions and stiff and slack tasks appear substantial, as they fall outside each other's standard deviation. This demonstrates that the setup allows for selectivity between different tasks and conditions. However, due to the small participant group, no conclusions can be drawn regarding its significance. The increased stiffness observed in the angled position compared to the neutral position may be attributed to the participant's effort to maintain the leg at that angle. Unlike the neutral position, where the leg is relaxed, in the angled position, the participant actively maintains the leg against gravity, while also resisting perturbations. This dual task may require additional muscular effort and contractions, which could result in increased stiffness.

4.2.1 NB versus WB

The FRFs of the NB trials converge toward those of the WB trials, as seen in Figure 12. This observation is consistent with findings from research with the Proprio perturbing the arm [12]. Therefore, the new setup still demonstrates this behaviour, featuring constant intrinsic and variable reflexive dynamics.

As mentioned earlier, two mechanisms are present to counteract external forces and to stabilise during posture control. The first mechanism involves intrinsic properties, which depend on the properties of the passive tissues and muscles. The second mechanism involves reflexive feedback, where muscle spindles provide sensory information on muscle stretch and stretch velocity and Golgi tendon organs on active muscle force [17].

In the NB trials, perturbations occur at a slower rate (below eigenfrequency), and reflexive feedback is active. Conversely, during the WB trials, force disturbances happen at slow rates, but also occur at high frequencies. With high frequencies, perturbations can only be resisted using intrinsic properties. Reflexive feedback is limited due to time delays associated with feedback from proprioceptive sensors [17].

Hence, the difference between NB and WB trials observed in Figure 12 could be a consequence of the modulation of reflex gains in the low-frequency NB trials, allowing for responses to perturbations. This conclusion can be further validated with parameter identification.

This reflex modulation was also a sensation reported by the participants. During trials with only low frequencies (< 2 Hz), participants felt they could actively respond to perturbations. However, with high-frequency disturbances, their strategy predominantly involved keeping their leg as stiff as possible.

4.2.2 Stiffness

Since stiffness is derived from the admittance functions, notable differences are observed. Specifically, larger differences have been found between neutral and angled positions compared to the right or left leg. Concerning the right and left legs, it remains unclear whether the lack of detectable differences is due to an actual absence of variation between the legs or limitations in the sensitivity of the setup or the Proprio system. It is not possible to compare differences between legs in the angled position, as this measurement was not conducted. Overall, the setup demonstrates selectivity when looking at stiffness values between different positions.

Validating stiffness values found in the current study presents challenges, primarily because leg stiffness has not been measured using this method. Conventionally, leg stiffness is assessed during activities such as running and jumping. However, these measurements may not be directly comparable to those obtained in the current study due to the significant methodological differences between the two approaches [36, 37].

A mechanical analysis could provide an assessment of stiffness for comprehension. Stiffness, denoted as k, is defined by the equation $k = \frac{F}{\Delta x}$, where F represents the applied load and Δx the resulting deformation [38]. Consider the free body diagram in Figure 13. It illustrates how the Proprio applies the disturbance force F_w and the resulting position deviations x. It shows that, given the leg is not fully in a vertical position in the angled stance, the effective force acting on the leg will be at an angle, which results in $F_{w,e}$. When using force decomposition, this results in $F_w > F_{w,e}$.

In this angled stance, it is assumed that the leg never reaches a position directly beneath the hip, resulting in $x < x_e$, where x_e represents the effective position deviation. The Proprio measures position deviation in the horizontal direction only, while the leg's movements encompass both horizontal and vertical components, forming an arc. This is why the free-body diagram of the leg is depicted as a pendulum.

Given that $F_w > F_{w,e}$ and $x < x_e$, it follows that $k_{neutral} > k_{angled}$. However, this is contradicted by the data, which shows that $k_{neutral} < k_{angled}$.

This discrepancy could arise from the assumption that maintaining the neutral position requires the same effort as the angled position. Although F_w is constant in both situations, the exerted force by the participants varies. The standard deviation of the exerted force measured indicates that $F_{exerted,neutral} < F_{exerted,angled}$, which says that the exerted force of the participant in the neutral position is smaller than that in the angled position (see Table 3). This applied to almost all participants, except for participant 7.

	Exerted force (standard deviation, N)	
Participant	Neutral position	Angled position
1	2.223	4.1123
3	2.4257	3.8595
5	2.5186	2.9164
6	2.7784	3.0551
7	4.0062	3.7365

Table 3: The standard deviation of the exerted force of the participants for the neutral and angled positions of the WB trials. The mean of the exerted force was 0.

The stiffness calculations used apply to elastic bodies. These have the property that material springs back to the original configuration. In the current context, the neutral and angled positions are the original configurations. The key difference is that maintaining the angled position requires muscle activation in the leg and hip to hold the position. Hence, the observed higher stiffness in the angle position can be attributed to the increased force needed to maintain the angled position. Consequently, even though the theoretical analysis suggests otherwise, the data confirms that stiffness is indeed greater in the angled position and it is important to consider muscle activation and exerted force when evaluating stiffness in biomechanical contexts.

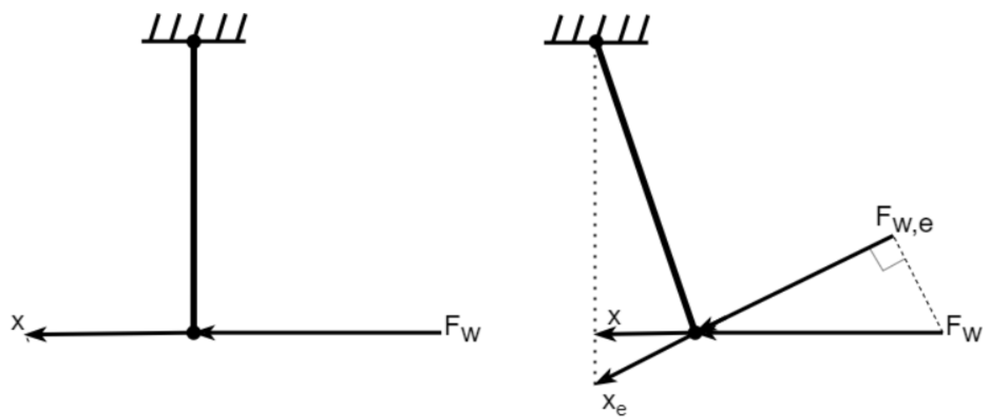

Figure 13: Free body diagram of the leg as a pendulum with applied force F_w working on the ankle (represented by the dot) and the resulting position deviation x with direction. $F_{w,e}$ and x_e are the effective force and position deviation in the working direction, thus perpendicular to the leg.

4.3 Comparing with arm response.

The objective of utilising the manipulator for the leg is to ultimately characterise the intrinsic and reflexive parameters and to further research the effect of fear of pain on performance. To achieve this, the outcomes obtained from the new setup must be comparable to those obtained previously. Van der Helm et al. (2002) devised a method to quantify reflexive feedback gains based on mechanical behaviour, as documented in their analysis of the FRFs for both the WB and NB trials on the arm [12].

Their findings reveal FRFs that exhibit similarities. Their functions closely resemble those of second-order systems. However, a notable difference is the lack of the eigenfrequency peak in the FRFs of the arm. This indicates that the arm system is critically damped. In contrast, the FRFs for the leg, for most participants, exhibit an eigenfrequency peak, suggesting underdamped behaviour. Underdamped systems, as observed in the leg, imply insufficient damping to prevent oscillations. While both critically damped and underdamped systems move quickly towards the equilibrium positions, an underdamped system moves more rapidly [39]. However, it also overshoots and exhibits oscillations around this equilibrium.

The current admittance value measured in this study is approximately around $10e^{-2}\ m/N$. In previous work focusing on the arm, admittance values around $10e^{-3}\ m/N$ have been reported [12, 18, 40]. Therefore, the admittance value for the leg is higher than those found in other works around the shoulder.

Admittance typically decreases with a reduced frequency bandwidth of the perturbations, co-contraction and reflexes [40]. However, since the bandwidth of the perturbations remains the same across studies, this cannot explain the higher admittance observed in the current study. The higher admittance observed may be attributed to reduced co-contraction and reflexes compared to other work. This suggests that posture maintenance in the current study may involve a higher level of compliance in comparison with shoulder posture maintenance.

A possible explanation for the difference in admittance could lie in the functional differences between the anatomical regions. The upper limbs are designed to be free, mobile and capable of intricate grasping movements. They possess a wide range of motion and can execute complex muscle movements. However, this mobility comes at the expense of stability. The shoulder joint, for instance, is highly mobile but inherently less stable [41]. To compensate for this, the upper limbs may require more co-contraction or reflexes to maintain stability and resist external perturbations. In contrast, the primary function of the lower limbs is to provide stability. The hip joint is inherently stable compared to the shoulder joint. Even without the support of soft tissues, the hip joint can remain intact, and dislocation requires a significant amount of force [41]. These functional differences could result in varying levels of co-contraction and reflexes required to maintain stability and resist external perturbations.

Regarding stiffness, their observations indicate values of approximately 1000 N/m for arm stiffness, which is higher than the leg stiffness that is currently observed. This indicates that the arm tends to be stiffer than the leg with this posture control task.

One possible explanation for the difference in stiffness between the arm and leg could lie in the degree of voluntary co-contraction. Stiffness increases with higher levels of co-contraction. Individuals may have a greater capacity for voluntary co-contraction in the arm muscles, given the proficiency in executing precise movements with the arm and the coordination associated with it [41].

Furthermore, they observed a phase progression from 0 to approximately -135 degrees for the admittance functions of both the NB and WB trials. In contrast, in the present study, the phase initiates at -45 degrees. Therefore, there is consistently a phase lag, with the output signal $x(t)$ lagging behind the input signal $w(t)$, which is not seen in the arm dynamics.

This may be due to the fact that the arm typically exhibits a lower signal time delay compared to the legs [42]. This is because the time delay is influenced by the distance signals need to travel through the body and the arm is closer to the spinal cord and the brain than the leg [42, 43]. Consequently, neural signals involved in reflexes and voluntary control in arm movements have a shorter distance to travel, which could result in a smaller phase lag.

4.4 Task association

The required task in this study presents a slight deviation from typical balance-related activities, potentially leading to a lack of intuitive association with everyday movements. Normally, individuals maintain balance by stabilising their entire body, whereas in this experiment, the focus is specifically on balancing the leg itself. This divergence from conventional tasks may diminish the compatibility of the experimental set-up to real-life scenarios. Furthermore, the task's requirement to activate hip muscles in a lateral direction is inherently less common compared to the activation patterns observed in the arms. In everyday movements, such as reaching or grasping objects, individuals frequently engage their arm muscles in various directions, whereas lateral co-activation of hip muscles is relatively infrequent. This discrepancy in muscle activation patterns further contributes to the perceived unnaturalness of the task. Consequently, the combination of the atypical task demands and the less frequent co-activation of hip muscles laterally may collectively contribute to the sensation of discomfort or unfamiliarity experienced by participants during the experiment.

Next to that, several participants noted that they felt they had better control using their dominant leg. They described feeling as though they could control their dominant leg more effectively, while with their non-dominant leg, they struggled to achieve the same level of control. Participants reported that their efforts to control their non-dominant leg seemed to have little effect. In contrast, they found it easier to maintain stiffness in their dominant leg. This observation was particularly notable among participants who regularly engage in sports that heavily emphasise the use of a dominant side, such as badminton.

4.5 Measuring muscle activity

Initially, the intention was to utilise electromyography (EMG) to measure muscle activity to facilitate better comparison among participants. Specifically, the muscles targeted for measurement were the Gluteus Medius, Adductor Magnus, and Gracilis. These muscles were selected due to their significant role in the abduction and adduction of the leg, which aligns with the direction of the perturbations in the experiment. However, the use of EMG measurements was ultimately abandoned due to privacy concerns and comfort. This decision stemmed from the discomfort that could potentially arise from the placement of electrodes, particularly considering that only a female researcher was available. For instance, for the Gluteus Medius, electrodes would need to be positioned relatively high up on the leg, while for the Adductor Magnus and Gracilis, landmarks on the inner thighs would need to be identified before electrode placement could occur in the appropriate locations [44].

5 Conclusion

In conclusion, this study aimed to evaluate the suitability of a new setup around the Proprio to measure postural control at the leg, employing the method by van der Helm, Vlugt and Schouten [12].

When wanting to evaluate the setup, requirements were compiled. First, the force disturbance must include the system's eigenfrequency, as this range of frequencies is crucial for postural control. The results indicate that the eigenfrequency was successfully excited, providing insight into the damping of the system and the changes in reflexive feedback. Second, observing the convergence of NB trial results to WB trial results is essential. This convergence has been observed and changes in reflexive dynamics can possibly be observed with the setup. Furthermore, identifying differences in FRFs and stiffness values across different trials is necessary. Differences were found between slack versus stiff conditions (in the FRF) and angled versus neutral positions (in the FRF and stiffness). The setup successfully measures different results, demonstrating its ability to differentiate between positions and trials. While estimating leg stiffness through spectral analysis was feasible, validating it was challenging due to methodological differences compared to existing literature. However, our findings were similar to previous studies, suggesting our study accurately measured leg dynamics. Finally, a high coherence is necessary to perform parameter identification in the future. The observed high coherence indicates that intrinsic parameters and reflex gains can be estimated through parameter identification. Considering these factors, the setup appears suitable for analysing lower extremity postural control using the Proprio.

Validating the results presents challenges given that it is the first study using this method at the lower extremity. The methodology established by van der Helm et al. provided a crucial reference point for comparison [12]. Notably, the leg demonstrates a tendency towards underdamped behaviour, contrasting with the arm's inclination towards critical damping. Moreover, higher admittance values in the leg compared to the arm imply different response mechanisms to perturbations and greater compliance in leg control tasks possibly due to reduced co-contraction and reflexes. Functional gaps between upper and lower limbs, stemming from differing mobility and stability requirements, are likely key contributors to these observed distinctions.

It is important to note that the task employed in this study deviates from a typical balance task, involving infrequent movements. This deviation may limit the compatibility of our setup with real-life scenarios and broader perspectives. Furthermore, the incorporation of EMG in future analyses will enable a comparison between participants to determine which muscles are more active, providing further insights into postural control mechanisms.

If it could be demonstrated that there is a change in neural or mechanical control due to fear of pain, rehabilitation strategies could be tailored accordingly. Depending on the situation, rehabilitation efforts could focus more on addressing either the mechanical or psychological aspects.

6 Future

This study marks the initial phase of system identification for the human leg using the Proprio, which is normally used for research around the shoulder joint and arm. As mentioned earlier, the integration of EMG measurements can enhance estimation accuracy and facilitate comparative analysis. The next step in this methodology involves expanding the participant group to ensure statistically significant results. Additionally, parameter identification can be conducted to determine intrinsic and reflexive parameters, allowing for a quantitative analysis of the system. Currently, system properties are only assessed through graphical representations. Once parameter identification is completed and validated, the impact of fear of pain on intrinsic and reflexive properties can be investigated using the Proprio and system identification techniques. This will enable a comprehensive study of the effect of fear of pain on movement control, extending beyond mere muscle activation and movement outcome analysis done in earlier research.

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