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TOWARDS OPTIMISATION OF PATELLAR TENDON LOADING IN MUSCULOSKELETAL REHABILITATION

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Chapter 1

General introduction

1.1 Mechanotherapy: rehabilitation exercises as medication

Loading is an important concept in the context of physical activity in general and musculoskeletal rehabilitation specifically. When applied to the human body, load is broadly defined as: "the sport and non-sport burden – single or multiple physiological, psychological or mechanical stressors – as a stimulus that is applied to a human biological system" (Soligard et al., 2016). Within this broad definition, a separation can be made between external and internal load. External load on the one hand refers to: "any external stimulus applied to the individual that is measured independently of their internal characteristics" (Halson, 2014; Soligard et al., 2016) such as running speed or the mass of external weights during squats. Internal load on the other hand refers to: "the load measurable by assessing internal response factors within the biological system" (Halson, 2014; Soligard et al., 2016). Examples of internal loads on the musculoskeletal system include objective measures like the forces produced by individual muscle-tendon units, but also more subjective measures like the rate of perceived muscle exertion (Vanrenterghem et al., 2017). Due to variations in biological and environmental factors, the same external load can result in different internal loads among individuals and among different structures within an individual (Soligard et al., 2016; Vanrenterghem et al., 2017).

The load capacity of a musculoskeletal tissue evolves throughout a person's lifetime due to mechanotransduction (Khan & Scott, 2009; Thompson et al., 2016). In brief, mechanotransduction describes a physiological process where cells embedded in the musculoskeletal tissues can sense the internal load that is imposed to them through mechanosensitive receptors in their membranes. Activation or deactivation of these mechanosensitive receptors results in a biochemical response that either promotes or inhibits the synthesis and remodelling of the tissue's extracellular matrix. In turn, the changes in the extracellular matrix will result in different internal forces that arrive at the level of the cells (Thompson et al., 2016). This feedback loop ensures that the load capacity of a musculoskeletal tissue is sufficient to cope with daily-life mechanical stressors. In the context of sports and training, the load capacity of a musculoskeletal structure can increase even further with repetitive bouts of overload and optimal recovery times: a phenomenon that is also known as supercompensation (Bourgois & Boone, 2016). However, the combination of overload and inadequate recovery can lead to overuse injuries (Dye, 2005; Hart & Scott, 2012; Windt & Gabbett, 2017). Fortunately, injured musculoskeletal structures can show positive adaptations in response to mechanical stimuli as well. Thus, therapeutic exercise programs can be

prescribed to promote the healing process of injured structures. This concept is typically referred to as mechanotherapy (Khan & Scott, 2009; Thompson et al., 2016). Thanks to mechanotherapy, people can recover from their musculoskeletal overuse injuries with rehabilitation exercises as a low cost, easily accessible and individualizable treatment option.

An ideal mechanotherapeutic rehabilitation program for a musculoskeletal injury would conceptually look like Figure 1.1. In the initial phase, the load on the injured structure is reduced to values below the current load capacity to break the vicious circle of overuse. However, absolute rest should be avoided to prevent further reduction of the load capacity of the injured structure (Bleakley et al., 2012; Dubois & Esculier, 2020). Next, the load on the injured structure is gradually increased according to the principles of overload and supercompensation until habitual load levels can be tolerated again (Glasgow et al., 2015; Hart & Scott, 2012). Meanwhile, the load on the surrounding uninjured structures should be kept as habitual as possible to prevent general disuse and deconditioning (Harvey et al., 2018). Perhaps, the load on these uninjured structures should even increase to improve their load capacity as well. This recommendation is made based on two observations. First, the contribution of kinetic chain deficits as a causal factor for local overuse injuries is more and more recognised. For example, it is well established that abnormal hip mechanics can be a causal factor for knee overuse injuries like patellofemoral pain syndrome, iliotibial band syndrome or patellar tendinopathy (Powers, 2010). Second, patients can use compensation strategies to reduce the load on their injured structure during physical activities, but this means that the surrounding structures will bear more load (Roos et al., 2014). If these surrounding structures are not able to cope with this additional load, they may sustain overuse injuries themselves. Progressively increasing the load capacity of an injured structure while simultaneously maintaining or further increasing the load capacity of the surrounding uninjured structures, requires a well thought-out rehabilitation program (Glasgow et al., 2015). Therefore, the patient should be supervised throughout the rehabilitation process by an expert in the field of physical activity during injury: the physiotherapist (WorldPhysiotherapy, 2019).

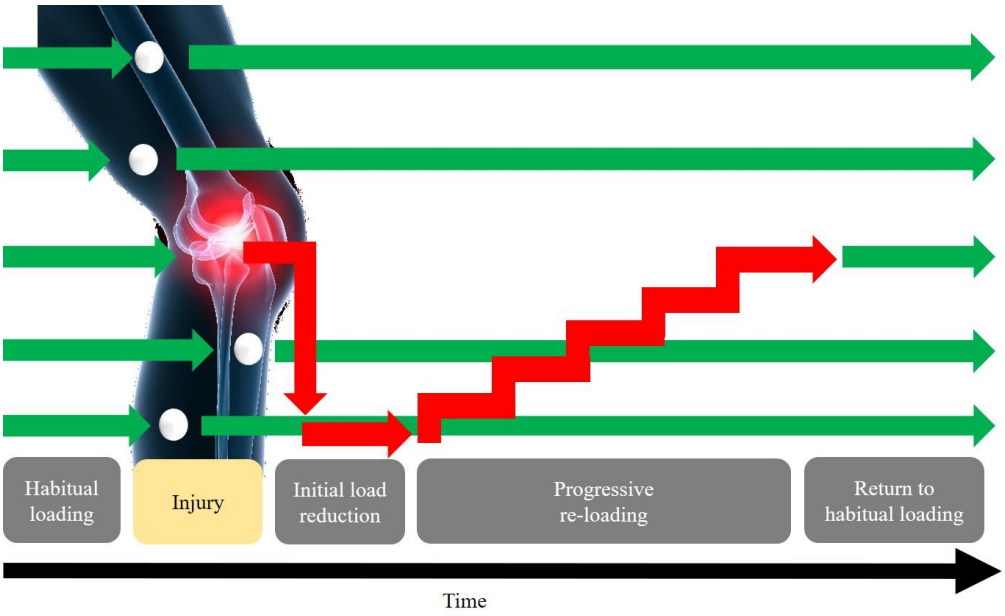

Figure 1.1. *Conceptual ideal scenario for mechanotherapy in which the loading of the healing tissue is 'uncoupled' from its surrounding tissues. After an injury, the load on the healing tissue is initially reduced, followed by progressive re-loading towards habitual loading levels (red arrow). For the surrounding tissues, loading is kept as habitual as possible to prevent unnecessary weakening or deconditioning (green arrows).*

1.2 Patellar tendinopathy: an exemplary overuse injury for applying mechanotherapy

Mechanotherapy can be used for the rehabilitation of overuse injuries in every type of musculoskeletal tissue (Khan & Scott, 2009). An exemplary overuse injury where mechanotherapy is described as a primary treatment intervention, is patellar tendinopathy (Cardoso et al., 2019). This overuse injury is defined as: "persistent patellar tendon pain and loss of function related to mechanical loading" (Scott et al., 2020). The patellar tendon is an elastic structure, mainly composed of collagen and proteoglycans, that connects the quadriceps muscle to the tibial bone through the patellofemoral joint. The primary function of the patellar tendon is to transfer the tensile forces generated by the quadriceps to the tibia (Thorpe & Screen, 2016). The patellofemoral joint assists in this function by acting as a pulley, changing the direction of the patellar tendon force and increasing its efficiency (Loudon, 2016). Furthermore, the elastic nature of the patellar tendon gives it a spring-like energy storage and release function (Thorpe & Screen, 2016). In result, patellar tendinopathy is a common injury in athletes practicing sports that involve such energy storage and release activities like running and jumping. A recent systematic review identified an overall patellar tendinopathy prevalence of 18.3% in a general athlete population and prevalence rates of 20.8% and 24.8% in basketball and volleyball players specifically (Nutarelli et al., 2023). Patellar tendinopathy is also a

disabling injury with a reported reduced work ability of 16% and reduced sports performance of 55% (Vries et al., 2017).

The pathophysiology of patellar tendinopathy is a complex interaction between loading on the one hand and structural abnormalities, pain and functional impairments on the other hand (Figure 1.2) (Cook et al., 2016; Cook & Purdam, 2009). In essence, overload and inadequate recovery of the patellar tendon cause a biochemical response of increased proteoglycan synthesis, collagen disorganisation and increased vascular and neuronal ingrowth. Appropriate loading and recovery times can reverse this process, but the more the tendon has evolved towards the degenerative stage, the less likely the structural integrity of the patellar tendon will completely recover (Cook & Purdam, 2009). Patellar tendon pain is also related to mechanical loading of the tendon. However, the pain can arise both in the presence or absence of macroscopic structural abnormalities. While several hypotheses have been formulated, the driver behind this patellar tendon pain remains unknown until now (Cook et al., 2016). Given the partial uncoupling between pain and structural abnormalities, patellar tendinopathy is primarily diagnosed based on clinical examination with medical imaging being used as an auxiliary tool in case of doubt (Malliaras et al., 2015; Scott et al., 2020). Finally, patellar tendon pain and/or structural abnormalities can be associated with impaired functionality, such as a reduced load capacity of the patellar tendon or strength deficits of the quadriceps muscle. Recovering these functional impairments – even in the absence of patellar tendon pain or structural abnormalities – is an important part of the rehabilitation process to reduce disability and prevent re-injury (Cook et al., 2016).

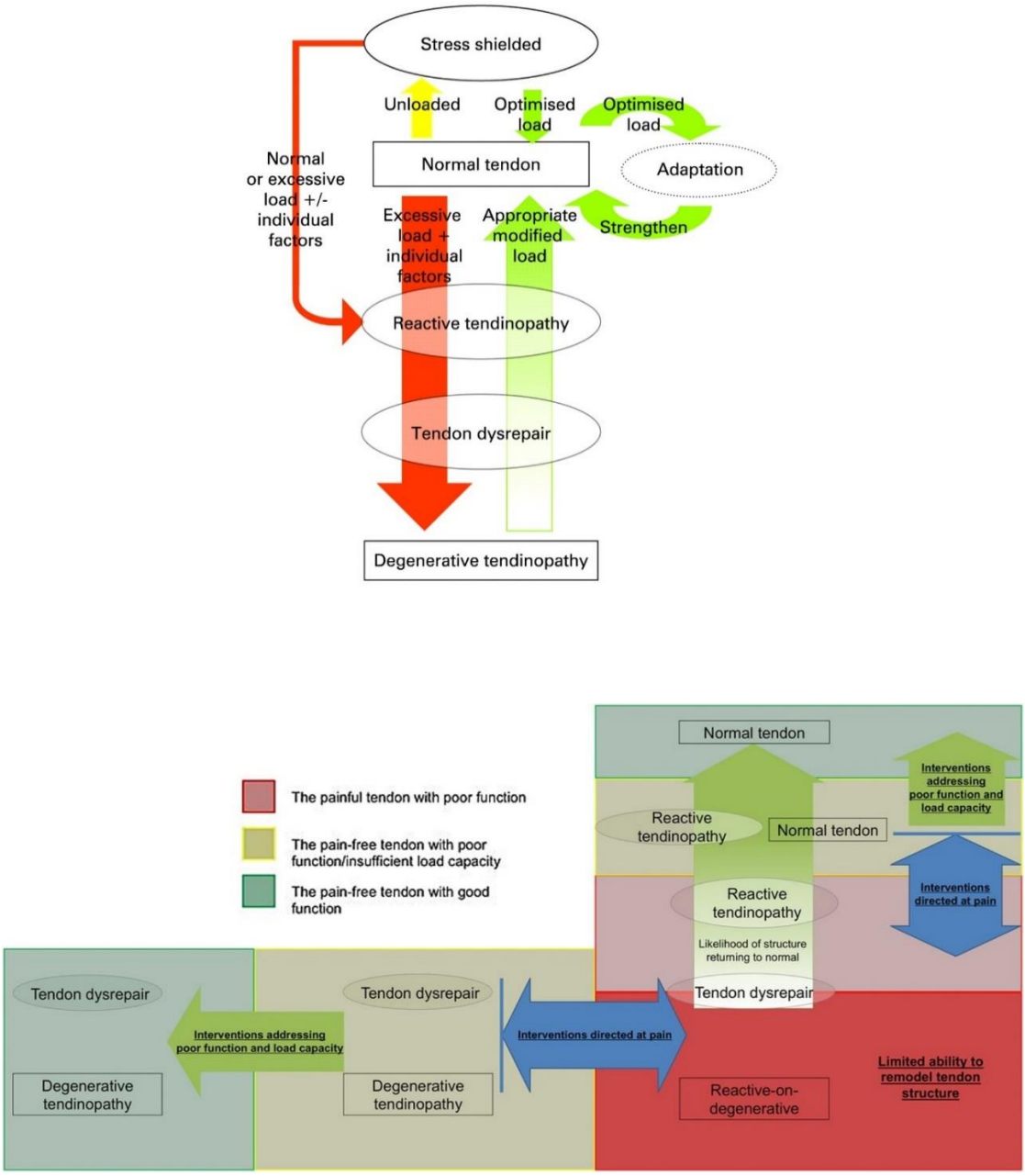

Figure 1.2. *Top: original tendon pathology continuum model (Cook & Purdam, 2009). The model explains the transition from a normal to a degenerative tendon and its relationship with loading. Bottom: updated tenon pathology continuum model (Cook et al., 2016). This updated version pays more attention to the complex interaction between structural, functional and pain aspects related to tendinopathy. Permission to use the figures was granted by BMJ.*

For long, it was thought that eccentric strength training was the optimal loading form for tendons, including the patellar tendon. This was because the first successful prospective intervention study for the conservative rehabilitation of Achilles tendinopathy by Alfredson et al. (1998) included an exercise program of heavy-load eccentric calf muscle exercises. A few years later, the same research group demonstrated similar positive effects of eccentric quadriceps exercises for the rehabilitation of patellar tendinopathy (Jonsson & Alfredson, 2005). However, the superior effects of accentuated eccentric exercise programs have been challenged more and more over the last years. First, Kongsgaard et al. (2009) showed that a heavy slow resistance protocol without eccentric accentuation had similar results as an accentuated eccentric exercise program in terms of pain reduction and structural improvement, but with a higher patient satisfaction. Next, Rio et al. (2015) provided evidence for a short-term analgesic effect through inhibitory neural pathways after isometric quadriceps contractions at 70% of the one repetition maximum. Finally, a recent systematic review showed that load intensity is a more important parameter for structural adaptations than contraction type (Lazarczuk et al., 2022). High intensity exercises had superior results compared to low intensity exercises, but there was no difference between concentric and eccentric exercise programs. The evidence mentioned above, inspired the creation of a clinical guideline presenting a progressive tendon loading program of four stages (Malliaras et al., 2015):

1. Isometric loading for pain reduction
2. Isotonic concentric-eccentric loading with progressively increasing intensity
3. Progressive re-introduction of energy storage and release loading
4. Gradual return to sport

This progressive tendon loading program has been tested against an accentuated eccentric exercise program and has shown superior results for pain reduction after 24 weeks (Breda et al., 2021). In conclusion, a progressive tendon loading program is the current standard of care for the rehabilitation of patellar tendinopathy (van Berkel et al., 2022).

1.3 Current best practice for progressive patellar tendon loading

During stage two and three of the progressive tendon loading program a gradual increase of the patellar tendon load – both between and within the stages – is key for a successful rehabilitation of patellar tendinopathy (Cook & Docking, 2015). For example, during the isotonic loading stage the mass of external weights will gradually increase, while the number of repetitions per exercise will gradually decrease according to the basic principles of strength training (Kongsgaard et al., 2009; Malliaras et al., 2015; Mc Ardle et al., 2014b; Vrijens & Bourgois, 2016). Similarly, the intensity and repetitions can be modified during the progressive re-introduction of energy storage and release exercises like running or jumping. While this is a logical approach at first sight, it should be noted that load progressions currently happen based on external load parameters instead of internal, patellar tendon specific load parameters. And again, the same external load can lead to different internal loads between individuals and between structures within an individual (Soligard et al., 2016; Vanrenterghem et al., 2017). An interesting example to support this statement, comes from the recent study by Armstrong et al. (2022), investigating muscle activations during traditional and eccentrically accentuated squats. While the external resistance increased from 20% to 150% of the one repetition maximum, the activity of the gluteus maximus did not further increase after 100% of the one repetition maximum and the activity of the vastus lateralis even reached a plateau at 80%.

At present, internal patellar tendon loads are subjectively measured through pain during the rehabilitation of patellar tendinopathy. The guidelines suggest that mild pain scores up to three on a ten-point numeric rating scale during or immediately after the exercise program are acceptable and even necessary to create an adaptive overload stimulus (Escriche-Escuder et al., 2020; Jonsson & Alfredson, 2005; Malliaras et al., 2015). However, the pain score on a loading test such as the single-leg decline squat should return to its baseline value within 24 hours after the end of the exercise program. If the pain has not returned to its baseline value, the load capacity has been exceeded too much and the external loads of the exercise program should be reduced (Escriche-Escuder et al., 2020; Malliaras et al., 2015). Using subjective measures, such as pain, as internal load parameters has its downsides though. First, psychosocial factors can be dominant contributors to patellar tendon pain, especially in chronic patellar tendinopathy patients (Edgar et al., 2022; Hanlon et al., 2021). In this subgroup of patellar tendinopathy patients with dominant psychosocial contributors, the relationship between tendon pain and tendon overload is partially or completely uncoupled. Second, mistakes in progressive patellar tendon loading are only detected when the pain score on the loading test does not return to its

baseline value. However, days or even weeks could expire between the start of the incorrect tendon loading and the increase of pain scores on the loading test (Escriche-Escuder et al., 2020; Magnusson et al., 2010). More objective internal patellar tendon load parameters could overcome these limitations.

An operational, biomechanical definition for internal patellar tendon load can be found by looking at a tendon's stress-strain relationship (Figure 1.3) (Wang, 2006). Here, we can appreciate that a certain amount of tensile force on the tendon divided by the tendon's cross-sectional area results in a certain amount of strain and potentially damage. Therefore, the maximum amount of tensile patellar tendon force (PTF) during a rehabilitation exercise can be considered the biomechanical internal patellar tendon load. However, the peak PTF is not the only relevant internal load parameter. Due to mechanical phenomena like creep, stress-relaxation and hysteresis, the stress-strain relationship of the patellar tendon is influenced by the contraction velocity and the time under tension (Kelc et al., 2013; Kösters et al., 2014; Pearson et al., 2007). Hence, the rate of PTF development – the speed with which the PTF increases – and the PTF impulse – the total amount of PTF over the time under tension – are also relevant load parameters (Benson et al., 2021; Firminger et al., 2020; Tayfur et al., 2022). The relationship between the PTF and the resulting damage is moderated by structural load capacity parameters like the patellar tendon cross-sectional area or the patellar tendon material properties that determine its stiffness (Figure 1.4).

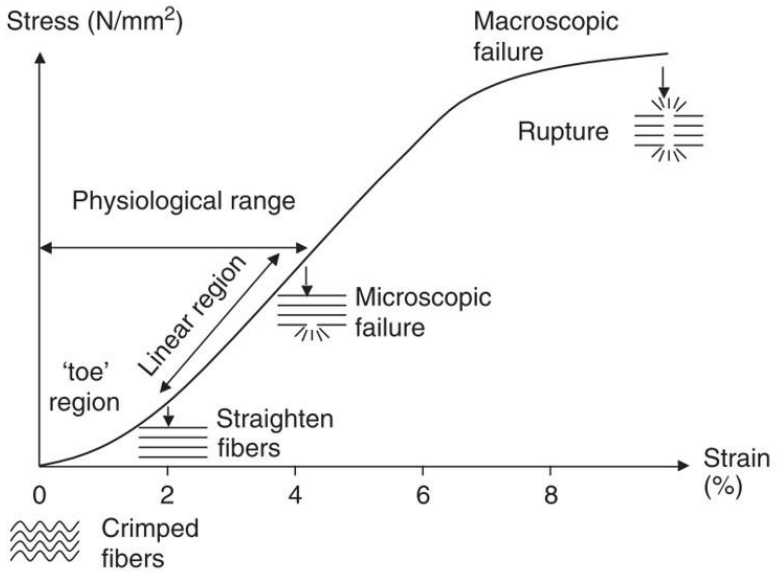

Figure 1.3. *Graphical representation of the stress-strain relationship in tendons* (Wang, 2006). *Tendon stress is the tendon force divided by the cross-sectional area of the tendon. Strain is the change in tendon length, represented as a percentage of the resting length. Permission to use the figure was granted by Elsevier.*

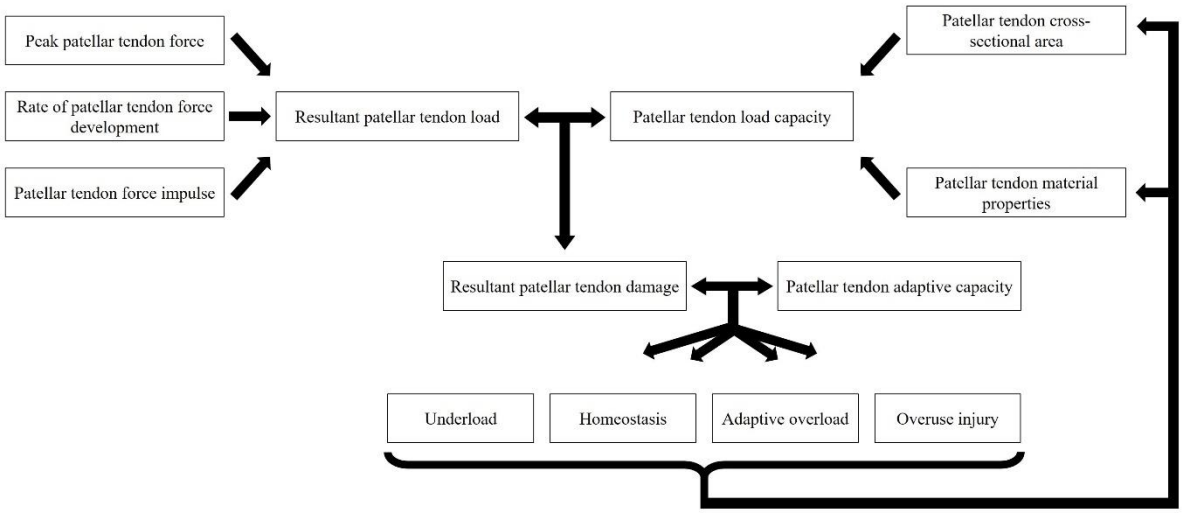

Figure 1.4. *Schematic model of the relationship between patellar tendon load and adaptation in the patellar tendon load capacity. The balance between the resultant patellar tendon load and the load capacity will lead to a certain amount of patellar tendon damage. Depending on the patellar tendon's adaptive capacity, the damage will be perceived as underload, homeostasis, adaptive overload or an overuse injury. This will have its effect on the load capacity.*

Unfortunately, measurement tools that can objectively measure PTF during rehabilitation exercises in a clinical practice setting are not available at this moment (Verheul et al., 2020). First of all, there is currently no measurement tool that can directly measure PTF in vivo in a non-invasive way. A buckle transducer or optic fibre sensor can measure PTF directly, but both require surgical implantation of the device (Fleming & Beynnon, 2004; Ravary et al., 2004; Roriz et al., 2014). In consequence, only two studies have directly measured PTF in vivo for a limited number of individuals and exercises. First, Finni et al. (2000) observed larger peak PTF values during a countermovement jump compared to a squat jump at moderate intensity. Second, Elvin et al. (2009) concluded that there were no significant differences in the peak PTF during the push-off and landing of a countermovement jump and that there was a moderate correlation between the peak PTF and the jump height.

Another way to objectively quantify PTF in vivo, would be to derive them from measured knee extensor moments. This would be possible through advanced musculoskeletal modelling and its accompanying methods to calculate the forces on individual musculoskeletal structures (Delp et al., 1990). However, the number of studies and exercises that have objectively quantified PTF with advanced musculoskeletal modelling is limited as well. To our knowledge, merely three studies have objectively quantified PTF using advanced musculoskeletal modelling. First, Lenhart et al. (2014) showed that increasing the step frequency decreased peak PTF while running at a constant running speed. Next, Zellmer et al. (2019) observed increased

peak patellar tendon stresses – i.e. PTF divided by the cross-sectional area of the patellar tendon – during a forward lunge when the knee was in front of the toes compared to behind the toes. Finally, Garcia et al. (2022) demonstrated that a forward jump landing generated a larger peak PTF than a vertical jump landing.

Unfortunately, advanced musculoskeletal modelling can currently only be used in scientific research and clinical gait laboratories. The measurement of joint moments requires expensive equipment such as a three-dimensional motion capture system and synchronised force plates (Uhlrich et al., 2023). Furthermore, data processing through advanced musculoskeletal modelling takes a considerable amount of time and requires a certain level of expertise. If physiotherapists want to include exercises into their rehabilitation programs that have not yet been objectively quantified by scientific research, they have to rely on their biomechanical and clinical reasoning skills. Furthermore, physiotherapists have to use their visual inspection skills to spot individual variations in movement patterns that can be relevant to the PTF. Take for example the position of the knee compared to the foot in the sagittal plane during squats, lunges or vertical jumps (Zellmer et al., 2019). We can thus conclude, that PTF estimations in a physiotherapy practice setting today are still largely subjective.

1.4 How to optimise load management: examples from other rehabilitation programs

The rehabilitation of patellar tendinopathy would benefit from more objective PTF measurements that can be performed in a physiotherapy practice, but how can this be accomplished? We could take an example from a field of rehabilitation that has already made major advancements in field-based internal load measures over the last decades: cardiorespiratory rehabilitation. Fifty years ago, objective heart rate measurements could merely happen through electrocardiographic devices and reliable spirometry measures required large Douglas bags (Burke, 1998; Mc Ardle et al., 2014a). Physiotherapy practices at that time were limited to palpations of the radial or carotid arteries for their internal load measures. To assist physiotherapists, large databases with normative data for energy expenditures during physical activities were created, like the Compendium of Physical Activities (Ainsworth et al., 2011). Nowadays, internal load measurement tools like heart rate monitors, oxygen saturation sensors and even portable spirometers are available for physiotherapists at reasonable prices. Patients with cardiorespiratory conditions now only need to visit specialised centres for testing with an electrocardiogram and spirometer when a more detailed analysis is indicated.

When we look for exemplary pathways within the field of musculoskeletal rehabilitation and biomechanical loading, the rehabilitation of knee osteoarthritis comes to mind. Here, a dataset of in vivo measures of knee joint contact forces from instrumented knee implants in older adults has been created (Heinlein et al., 2007; Taylor et al., 2017). Advanced musculoskeletal models are typically validated against the data from the instrumented knee implants (Hicks et al., 2015; Pelegrinelli et al., 2023) and can in turn objectively quantify knee joint contact forces in an even wider variety of situations. Musculoskeletal models can on the one hand be used in scientific research to further expand the database of objectively quantified knee joint contact forces. For example, Van Rossom et al. (2018) have objectively quantified knee joint contact forces over a variety of rehabilitation exercises for a group of young individuals. On the other hand, musculoskeletal models can be used to analyse clinical gait lab data of patients with knee osteoarthritis to tailor their surgical interventions (Ruggeri et al., 2022). And already now, the first efforts are made towards model-informed rehabilitation interventions (Uhlrich et al., 2022) and real-time feedback on knee joint contact forces in a physiotherapy practice setting (Saliba et al., 2018).

Based on these exemplary pathways, the first two steps towards the optimisation of patellar tendon loading in musculoskeletal rehabilitation are:

1. The objective quantification of PTF in a laboratory setting with advanced data collection and processing techniques, eventually creating a sufficiently large database of objectively quantified PTF during rehabilitation exercises.
2. Creating measurement tools that can objectively quantify PTF in a physiotherapy practice setting, preferably in real time.

1.5 Objectives

From the literature overview above, it is clear that mechanotherapy is an important concept within the field of physiotherapy (Khan & Scott, 2009; Thompson et al., 2016). Patellar tendinopathy is one overuse injury where mechanotherapy is considered a primary component of the treatment intervention (Cardoso et al., 2019; van Berkel et al., 2022). Progressive tendon loading programs have been developed and tested in randomised controlled trials (Breda et al., 2021; Malliaras et al., 2015). However, these programs could be further improved through the objective quantification of PTF during rehabilitation exercises. Therefore, the overall aim of this doctoral thesis was to improve the objective quantification of PTF progressions in a physiotherapy practice setting (Figure 1.5). Meeting this overall objective involved two steps, each specified in a specific objective:

Objective 1: objectively quantify PTF progressions during frequently used exercises in the rehabilitation of patellar tendinopathy using state-of-the-art data collection and processing techniques.

The number of studies that have objectively quantified PTF using advanced musculoskeletal modelling has been limited up to now. This thesis expands upon the existing scientific literature by investigating additional PTF progressions. More specifically, the (interacting) effects of changes in multiple external load parameters were investigated.

Study 1: investigate the influence of changes in speed and step frequency during running (n = 12 healthy participants).

Study 2: investigate the influence of an external weight's mass and carrying position during a forward lunge (n = 10 healthy participants).

Study 3: investigate the influence of an external weight's mass and the use of a decline board during a heavy-load single-leg squat (n = 12 healthy participants).

Because of the close relationship between PTF and patellofemoral joint contact forces and because running and lunges are also relevant exercises for the rehabilitation of patellofemoral pain syndrome – another prevalent overuse injury in young, physically active individuals – patellofemoral joint loading was reported for these two exercises as well.

Study 1-3 investigated PTF progressions in healthy individuals. Other studies on the objective quantification of PTF progressions have typically included healthy individuals as well. While these studies are relevant for the prevention of patellar tendinopathy, it is important to check whether these PTF progressions can be used for the rehabilitation of patients with patellar tendinopathy as well. PTF progressions may be different in these patients due to pain-avoidance strategies for example. Therefore, it is important to verify if PTF progressions are similar in patients with patellar tendinopathy and healthy individuals.

> **Study 4**: assess to what extent PTF progressions over a large number of rehabilitation exercises are equivalent in patients with patellar tendinopathy and healthy individuals (n = 7 patients with patellar tendinopathy – 7 healthy participants).

Objective 2: taking initial steps towards measurement tools that can objectively quantify PTF in a physiotherapy practice setting.

At present, physiotherapists can only estimate PTF during rehabilitation exercises in a physiotherapy practice setting based on their visual inspection and clinical reasoning skills. However, the reliability of their force-estimating ability has never been investigated.

> **Study 5**: examine whether physiotherapists can accurately estimate PTF during rehabilitation exercises (n = 1 healthy participant – 100 physiotherapists).

An important step towards the development of clinical measurement tools, is the identification of proxy signals. These proxy signals are signals that vary in accordance with the actual load parameter of interest and can be measured with non-invasive, low-cost technologies.

> **Study 6**: take initial steps towards the identification of potential proxy signals for PTF over multiple rehabilitation exercises (n = 9 patients with patellar tendinopathy – 7 healthy participants).

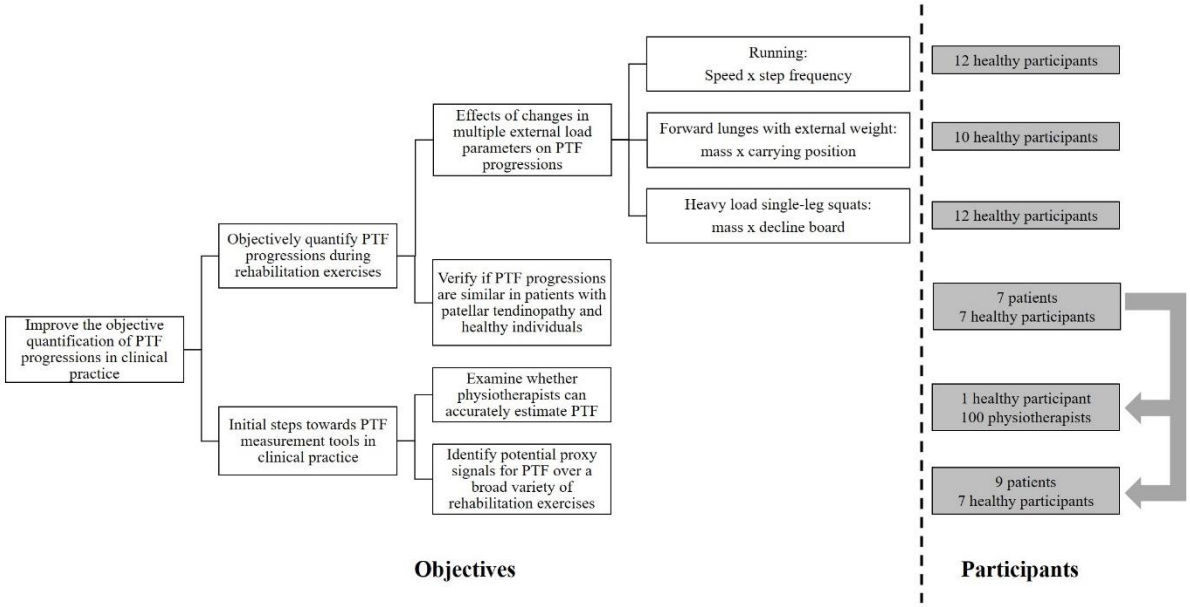

Figure 1.5. *Overview of different objectives in this doctoral thesis and the participant flow between the different studies.*

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Chapter 2

Study 1: Increasing step frequency reduces patellofemoral joint stress and patellar tendon force impulse more at low running speed

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Abstract

Objectives

Patellofemoral pain syndrome and patellar tendinopathy are important running-related overuse injuries. This study investigated the interaction of running speed and step frequency alterations on peak and cumulative patellofemoral joint stress (PFJS) and patellar tendon force (PTF) parameters.

Methods

Twelve healthy individuals completed an incremental running speed protocol on a treadmill at habitual, increased and decreased step frequency. Peak PFJS and PTF, peak rate of PFJS and PTF development and PFJS and PTF impulse per kilometre (km) were calculated using musculoskeletal modelling.

Results

With increasing running speed, peak PFJS ($p<0.001$) and PTF ($p<0.001$) and peak rate of PFJS ($p<0.001$) and PTF ($p<0.001$) development increased, while PFJS ($p<0.001$) and PTF ($p<0.001$) impulse per km decreased. While increasing step frequency by 10%, the peak PFJS ($p<0.001$) and PTF ($p<0.001$) and the PFJS ($p<0.001$) and PTF ($p<0.001$) impulse per km decreased. No significant effect of step frequency alteration was found for the peak rate of PFJS ($p=0.008$) and PTF ($p=0.213$) development. A significant interaction effect was found for PFJS ($p<0.001$) and PTF ($p<0.001$) impulse per km suggesting that step frequency alteration was more effective at low running speed.

Conclusion

The effectiveness of step frequency alteration on PFJS and PTF impulse per km is dependent on the running speed. With regard to peak PFJS and PTF, step frequency alteration is equally effective at low and high running speeds. Step frequency alteration was not effective for peak rate of PFJS and PTF development. These findings can assist the optimisation of patellofemoral joint and patellar tendon load management strategies.

Key words

Patellofemoral pain syndrome; patellar tendinopathy; load management; running mechanics

2.1 Introduction

Running has grown in popularity in the past six decades due to its accessibility, major health benefits and low cost (Lee et al., 2017; Scheerder et al., 2015). On the other hand, running poses a major injury risk, with injury prevalence numbers of 44.6% on average (Kakouris et al., 2021). More than 70% of these running-related injuries is associated with overuse (Kakouris et al., 2021). The knee joint is cited as the most common region for running-related injuries, with patellofemoral pain syndrome and patellar tendinopathy among the most prevalent ones (Benca et al., 2020; Francis et al., 2019; Kakouris et al., 2021). These overuse injuries occur when cumulative tissue microdamage exceeds the adaptive capacity of that tissue (Cook & Purdam, 2009; Dye, 2005). Load management is therefore an important aspect in the prevention and rehabilitation of both patellofemoral pain syndrome (Lack et al., 2018) and patellar tendinopathy (Malliaras et al., 2015).

Tissue microdamage during running is the result of a combination of the peak load per step and the cumulative load per kilometre (km) (Firminger et al., 2020). Biomechanical load parameters that are typically investigated for their association with overuse injuries are peak load and peak rate of load development per step and load impulse per km (Ceyssens et al., 2019; Firminger et al., 2020). An intervention that is often used to reduce these load parameters during running, is the alteration of the step frequency. Several studies have provided supportive evidence that an increased step frequency can reduce peak and impulse measures of patellofemoral joint stress (PFJS) and peak measures of patellar tendon force (PTF) for an individual step (Lenhart et al., 2014; Willson et al., 2014). During these studies, the running speed was kept constant over the different step frequency conditions. The effect of running speed itself has also gained increasing interest over the last years. Several studies suggest that an increase in running speed is accompanied by a decrease in impulse per km, because of the decreased number of steps per km and contact time per step (Petersen et al., 2015; Starbuck et al., 2021). Remarkably, the interaction effect of running speed and step frequency alteration on PFJS and PTF has not yet been investigated. In other words, whether the effect of step frequency alterations is different at high compared to low running speeds, remains unknown.

Therefore, the objective of this study was to investigate whether the load-modifying effect of step frequency alterations was similar at different running speeds. More specifically, the load parameters of interest were the peak PFJS and PTF per step, the peak rate of PFJS and PTF development per step, and the PFJS and PTF impulse per km running. We hypothesised that an interaction effect between running speed and step frequency alterations will be present for these three load parameters since the combination of high running speed and increased step frequency might trigger a shift in the running style, including the ratio between contact time and flight time, vertical excursion of the centre of mass, range of motion in the different joints and foot strike pattern (Allen et al., 2016; Dorn et al., 2012; Schubert et al., 2014).

2.2 Methods

2.2.1 Participants

Thirteen participants were recruited among students of the Faculty of Movement and Rehabilitation Sciences, KU Leuven (BEL). The participants were recreationally active runners without a lower limb injury or low back pain in the past six months and without a history of major lower limb injury that required surgery. No specific criterion was set in terms of minimum weekly running distance. An a-priori sample size calculation was performed in G*Power 3.1 (Erdfelder et al., 2009). For an arbitrary moderate effect size Cohen's f of 0.3 and an alpha of 0.008, 12 participants were necessary to detect significant interaction effects with a statistical power of 0.80 or higher. An alpha of 0.008 was used as a Bonferroni correction was applied (dividing an alpha value of 0.05 by the number of load parameters, i.e. six). The study was approved by the ethical committee of research UZ/KU Leuven (reference number: S62706). All participants provided written informed consent before the start of the data collection.

2.2.2 Protocol

The running protocol was performed on an instrumented treadmill (Motek Medical, Amsterdam, NL). All participants wore the same type of shoe (Indoor Copa, Kelme, Elche, ESP). A five-minute warm-up run at a self-selected speed was performed before the beginning of the data collection. Next, the participants completed three repetitions of an incremental speed protocol. This protocol started at an initial running speed of 8km/h and increased by 2km/h every minute until the speed of 16km/h was reached. During the first repetition, participants ran at their preferred step frequency. The step frequency per minute was determined at each running speed by counting the number of contacts from one foot during a 30-second interval and

multiplying this number by four (Allen et al., 2016; Heiderscheit et al., 2011; Thakkar et al., 2019). During the second and third repetition of the protocol, the participants were instructed to either increase or decrease their step frequency by 10% according to audible cues of a metronome. The order of the altered step frequency conditions was randomised for each participant using a coin toss. An eight-minute resting period in between repetitions of the protocol was provided to avoid fatigue effects. This could, however, be extended if the participant self-indicated that more rest was needed. In order to verify whether systematic fatigue effects were present, the rate of perceived muscle exertion was monitored at the end of each running speed block using a CR100 scale (Borg & Borg, 2001). The participant had to provide a fatigue score between 0: "nothing at all", and 120: "absolute maximum". The results can be found in supplementary analysis A.

2.2.3 Data collection

Kinematics of the lower limbs and trunk were collected at a sampling frequency of 100Hz by 13 cameras of a passive marker-based motion capture system (Vicon, Oxford, UK). Reflective markers were placed according to the Liverpool John Moores University model (Malfait et al., 2014). Ground reaction forces (GRF) were measured at a sampling frequency of 1000Hz using two force plates embedded in the instrumented treadmill (Aarts et al., 2018). Marker and GRF data were collected over a 15-second time interval during the steady state phase of each running speed block. That steady state phase was defined by two characteristics. First, the trunk of the participant did not make any horizontal motion relative to the treadmill, indicating that the participant was running at the desired running speed. Second, the participant had to match the step frequency indicated by the metronome. Both aspects were visually verified by two of the researchers who were present during the protocol (Chumanov et al., 2012; Heiderscheit et al., 2011). Data collection only started when both researchers confirmed these criteria have been met.

2.2.4 Data processing

The stance phases of 10 valid steps for one leg were retained for further analysis. The choice of which leg would be analysed, was randomised by a coin toss. Marker and GRF data were low-pass filtered with a cut-off frequency of 18Hz. A 20N threshold of the vertical component of the GRF marked the start and end of the stance phase.

Musculoskeletal modelling was performed in OpenSim 3.3 (OpenSim, Stanford, CA, USA) (Delp et al., 2007). The Catelli model was used as generic musculoskeletal model for this study because it has its muscle properties based on young and active individuals and it includes a patellofemoral joint and patellar tendon (Rajagopal et al., 2016). It also includes refined knee joint improvements capable of simulating fast running (Lai et al., 2017), while controlling the muscle paths during large ranges of the knee and hip joints (Catelli et al., 2019). This full-body model consists of 37 degrees of freedom (DoF). The lower limbs of this model consisted of a 6 DoF pelvis, a 3 DoF ball-and-socket hip joint, a 1 DoF tibiofemoral joint, a 1DoF patellofemoral joint, and a 1 DoF ankle joint. The motions of the lower limb were driven by 80 Hill-type muscle-tendon units. At the level of the quadriceps, the muscle-tendon units run all the way to the tibial tuberosity, and therefore include the patellar tendon. In other words, the patellar tendon is not modelled as a separate passive elastic element, but as an integral part of the muscle-tendon unit. The force transmission mechanism between the quadriceps and patellar tendon was modelled through fixed path points at the superior and inferior border of the patella (Rajagopal et al., 2016). Motions of the 6 DoF trunk segment were driven by ideal torque actuators. Since no marker data from the arms were collected, the upper limb segments of the model were fixed in a typical running position: 0° of shoulder flexion, adduction, and internal rotation and 90° of elbow flexion and pronation. The spatial dimensions of the segments, the optimal muscle fibre lengths and tendon slack lengths in the generic model were scaled to the values of the individual participants using the marker data from a static calibration trial. The maximum isometric muscle forces were multiplied by three to avoid major reliance on reserve torque actuators during the fast concentric contractions that take place during running at high speed (Kotsifaki et al., 2022; Swinnen et al., 2019). Joint kinematics during running were determined using a global optimisation approach, minimising the weighted sum of squared differences between the measured marker positions and the marker positions of the model (Lu & O'Connor, 1999). The movement of the patella was controlled by a spline function with the tibiofemoral flexion angle as a determinative variable. Net joint moments were calculated based on the kinematics results and the GRF data using the inverse dynamics tool. Forces of individual muscle-tendon units were calculated with a static optimisation approach (Mokhtarzadeh et al.,

2014). In this approach, the sum of the moments produced by the individual muscle-tendon units has to match the net joint moment for each frame and the sum of all muscle activations squared has to be minimal. To determine these muscle activations, the force-length and force-velocity relationships of the muscle-tendon units were taken into account. Finally, the kinematics, GRF, and muscle-tendon forces data were used to calculate joint reaction forces in the OpenSim joint reaction analysis tool (Steele et al., 2012).

Post-modelling processing was performed in Matlab R2020b (The MathWorks, Natick, MA, USA). The patellofemoral joint contact force was defined as the part of the total patellofemoral joint reaction force perpendicular to the patellofemoral joint surface. The PFJS was then calculated by dividing the patellofemoral joint contact force by the angle-specific patellofemoral joint contact area. This angle-specific contact area was determined using sex-specific regression functions created by Starbuck et al. (2021) (Starbuck et al., 2021) based on the MRI data of Besier et al. (2005) (Besier et al., 2005). PFJS values were expressed in megapascal (MPa). The PTF was defined as the force with which the patellar tendon pulls at the tibial tuberosity and was calculated as the scalar sum of forces produced by the four quadriceps muscle-tendon units. This simplification was possible since all muscle-tendon units of the quadriceps run in parallel at the level of the patellar tendon and insert at the tibial tuberosity. PTF was expressed in relation to the participants' body weight (BW). For each step, the peak PFJS and PTF were determined. Next, the peak rate of PFJS and PTF development and the PFJS and PTF impulse per step were determined using numerical differentiation and integration, respectively, of the PFJS-time curve and PTF-time curve. Finally, the PFJS and PTF impulse per km were calculated by multiplying the PFJS and PTF per step by the number of strides per km (Willy et al., 2016).

2.2.5 Statistical analysis

Peak load, peak rate of load development and load impulse per km were implemented in the statistical analysis. This statistical analysis was performed in the open-source software package SPM1D M0.4.9 (www.spm1d.org) (Pataky, 2012). All load parameters investigated in this study were zero-dimensional. Two-way repeated measures ANOVA's were performed to assess the effect of running speed, step frequency and their interaction. Since normality tests showed that all parameters were not normally distributed, the non-parametric version of the repeated measures ANOVA based on the principle of permutation testing was chosen (Nichols & Holmes, 2003). Note, the a-priori sample size calculation was made under the assumption of a normal distribution of the data and consequentially parametric statistics. However, non-

parametric permutation testing can be as powerful or sometimes even more powerful than parametric testing (Nichols & Holmes, 2003). The alpha value was set to 0.008.

In case of a significant effect in the repeated measures ANOVA, linear regression analyses were chosen as post-hoc analyses. The traditional paired t-test was not chosen as post-hoc analysis since a significant interaction effect in one parameter would result in a total of 105 paired t-tests. For each step frequency condition, a separate linear regression equation was generated with running speed as independent variable and the load parameter of interest as dependent variable. The regression coefficients represented the amount of increase or decrease in the load parameter with every 1km/h increase in running speed. In case of a significant main effect for running speed, the regression coefficients would be significantly different from zero, i.e. a horizontal line. The regression intercepts represented the values of the load parameter for the different step frequency conditions at a theoretical running speed of 0km/h. In case of a significant main effect for step frequency without interaction effect, the regression lines would be in parallel and the differences between the step frequency conditions, represented by the regression intercept, would be equal at every running speed. In case of a significant interaction effect, the regression coefficients would significantly different from one another and the regression lines would no longer be in parallel. Therefore, the differences in the load parameter between the step frequency conditions would be different at each running speed.

2.3 Results

The data of one of the 13 participants were withdrawn from the statistical analysis because of technical issues during the data collection. The resulting 12 participants were 6 men and 6 women aged 22 ± 3 years, with a body mass of 68.5 ± 6.7kg, and height of 1.77 ± 0.10m (mean ± standard deviation). Absolute and relative values for the step frequencies at the different running speeds can be found in Table 2.1. The procedure aimed to increase or decrease the step frequency by 10%. Therefore, each relative frequency should ideally be 10 ± 0%. However, some variation in the step frequencies is visible. Next to the variation for each running speed, it is also important to know the within-subject variation in the context of a repeated measures design. The differences between the maximum and minimum step frequency were 4 ± 2% in both the increased and decreased condition. Information about other spatial-temporal parameters can be found in supplementary analysis B.

	Decreased frequency	Habitual frequency	Increased frequency
8km/h	145 ± 7 (-8 ± 3)	157 ± 7	173 ± 10 (10 ± 3)
10km/h	150 ± 7 (-7 ± 3)	161 ± 8	179 ± 9 (11 ± 2)
12km/h	155 ± 8 (-6 ± 4)	166 ± 9	184 ± 10 (11 ± 2)
14km/h	161 ± 10 (-6 ± 4)	171 ± 10	189 ± 13 (11 ± 3)
16km/h	169 ± 12 (-5 ± 3)	178 ± 12	196 ± 15 (10 ± 4)

Table 2.1. *Mean values ± standard deviations* for the step frequencies at the different running conditions. The top values represent absolute values in steps per minute. The bottom values between brackets in the decreased and increased frequency condition represent relative percentages compared to the habitual frequency condition at the same running speed.

2.3.1 Patellofemoral joint stress

Descriptive values for the PFJS load parameters can be found in Figure 2.1 and Table 2.2. For the peak PFJS, significant main effects were confirmed for running speed ($p<0.001$) and step frequency ($p<0.001$), with higher running speeds resulting in higher joint stress and higher step frequency in lower joint stress. There was no statistically significant interaction effect ($p=0.338$). The regression analysis of the peak PFJS showed separated parallel lines that increased similarly with increasing running speed. The regression coefficient for the habitual step frequency was 0.48MPa. For both the increased and decreased step frequency conditions, the regression coefficient was 0.38MPa. The increased step frequency condition decreased the peak PFJS, and conversely, the decreased step frequency condition increased the peak PFJS. The regression intercepts for habitual step frequency, increased step frequency and decreased step frequency were 3.65MPa, 2.96MPa and 5.85MPa, respectively.

The peak rate of PFJS development showed a significant main effect for running speed ($p<0.001$), a main effect for step frequency that tended towards statistical significance ($p=0.008$), but no significant interaction effect ($p=0.147$). The regression coefficients were 14.05MPa/s for habitual step frequency, 8.66MPa/s for increased step frequency and 15.40MPa/s for decreased step frequency.

For the PFJS impulse per km, significant main effects for running speed ($p<0.001$) and step frequency ($p<0.001$), and a significant interaction effect ($p<0.001$), were found. The regression analysis showed decreasing and converging lines as the running speed increased. The regression coefficients were −31.17MPa*s/km for habitual step frequency, −22.36MPa*s/km for increased step frequency and −36.86MPa*s/km for decreased step frequency. An increase in step frequency decreased the PFJS impulse per km and a decrease in step frequency increased the PFJS impulse per km. The regression intercepts were 796.25MPa*s/km for habitual step frequency, 635.35MPa*s/km for increased step frequency and 908.84MPa*s/km for decreased step frequency. However, the effect of step frequency alteration was larger at lower velocities compared to higher velocities.

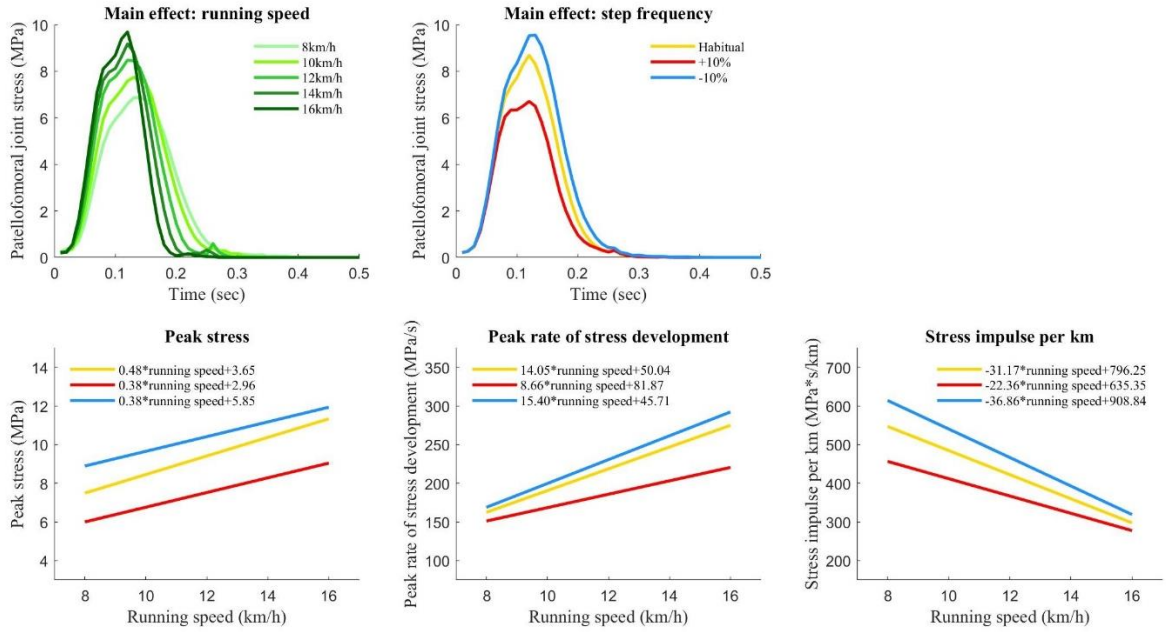

Figure 2.1. *Upper row: mean values of the PFJS -expressed in megapascal (MPa)- over time. Values are grouped according to running speed, combining all step frequencies (left) and according to step frequency, combining all running speeds (right). Lower row: regression lines peak PFJS (left), peak rate of PFJS development (middle) and PFJS impulse per kilometre (right). Red lines represent the increased step frequency condition, yellow lines the habitual step frequency condition and blue lines the decreased step frequency condition.*

Peak PFJS

	Mean ± SD values (MPa)		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
8km/h	8.63 ± 2.83	7.42 ± 2.21	5.97 ± 2.26
10km/h	9.86 ± 2.72	8.41 ± 2.67	6.92 ± 2.41
12km/h	10.54 ± 2.93	9.33 ± 2.63	7.56 ± 2.76
14km/h	11.28 ± 3.77	10.43 ± 2.78	8.25 ± 3.14
16km/h	11.72 ± 3.29	11.16 ± 3.31	9.15 ± 3.27
	Two-way repeated measures ANOVA		
	Running speed	**Step frequency**	**Speed*Frequency**
F-value	23.213	35.610	1.153
p-value	<0.001	<0.001	0.338
	Regression analysis		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
Coefficient	0.38	0.48	0.38
Intercept	5.85	3.65	2.96

Peak rate of PFJS development

	Mean ± SD values (MPa/s)		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
8km/h	155.94 ± 40.79	147.55 ± 36.60	139.76 ± 36.27
10km/h	210.57 ± 72.53	192.62 ± 60.89	180.57 ± 54.95
12km/h	236.15 ± 83.50	241.21 ± 99.48	189.51 ± 65.66
14km/h	269.23 ± 113.07	255.09 ± 92.42	204.57 ± 61.59
16km/h	280.59 ± 84.49	256.83 ± 83.86	214.32 ± 58.77
	Two-way repeated measures ANOVA		
	Running speed	**Step frequency**	**Speed*Frequency**
F-value	14.857	5.797	1.558
p-value	<0.001	0.008	0.147
	Regression analysis		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
Coefficient	15.40	14.05	8.66
Intercept	45.71	50.04	81.87

PFJS impulse per km

	Mean ± SD values (MPa*s/km)		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
8km/h	624.47 ± 215.21	563.11 ± 173.69	461.71 ± 179.04
10km/h	528.12 ± 169.76	470.14 ± 156.17	406.25 ± 150.28
12km/h	466.03 ± 112.12	411.02 ± 125.17	366.64 ± 122.04
14km/h	388.01 ± 98.54	361.11 ± 95.56	318.87 ± 102.12
16km/h	325.91 ± 78.20	305.97 ± 82.66	281.82 ± 80.78
	Two-way repeated measures ANOVA		
	Running speed	**Step frequency**	**Speed*Frequency**
F-value	34.951	53.670	7.490
p-value	<0.001	<0.001	<0.001
	Regression analysis		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
Coefficient	-36.86	-31.17	-22.36
Intercept	908.84	796.25	635.35

Table 2.2. *Mean values ± standard deviations and results of the two-way repeated measures ANOVA and regression analyses for PFJS load parameters.*

2.3.2 Patellar tendon force

Descriptive values for the PTF load parameters can be found in Figure 2.2 and Table 2.3. Like the peak PFJS, the peak PTF revealed statistically significant main effects for running speed ($p<0.001$) and step frequency ($p<0.001$) without a significant interaction effect ($p=0.376$). Peak PTF increased with increasing running speed. The regression coefficients were 0.37BW for the habitual step frequency, 0.31BW for the increased step frequency and 0.30BW for the decreased step frequency. Regression intercepts altered between 2.13BW for the increased step frequency condition and 4.60BW for the decreased step frequency condition. The peak rate of PTF development showed a significant main effect for running speed ($p<0.001$). Peak rates of PTF development increased as running speed increased with the regression coefficient altering between 11.90BW/s and 15.04BW/s. In contrast to the PFJS, the peak rate of PTF development showed no main effect for step frequency ($p=0.213$). The interaction of running speed and step frequency was not statistically significant for the peak rate of PTF development ($p=0.354$). The regression lines were close together and increased with increasing running speeds. The PTF

impulse per km showed a significant main effect for running speed ($p<0.001$), a significant main effect for step frequency ($p<0.001$) and a significant interaction effect between both factors ($p<0.001$). The PTF impulse per km decreased with increasing running speed. The regression coefficients were −24.03BW*s/km for habitual step frequency, −16.41BW*s/km for increased step frequency and −28.31BW*s/km for decreased step frequency. At low running speeds, the different step frequencies were separated. The highest regression intercept was detected at the decreased step frequency condition: 707.93BW*s/km. The lowest regression intercept of 485.21BW*s/km was detected at the increased step frequency condition. Towards 16km/h, the regression lines showed barely any difference between the step frequency conditions.

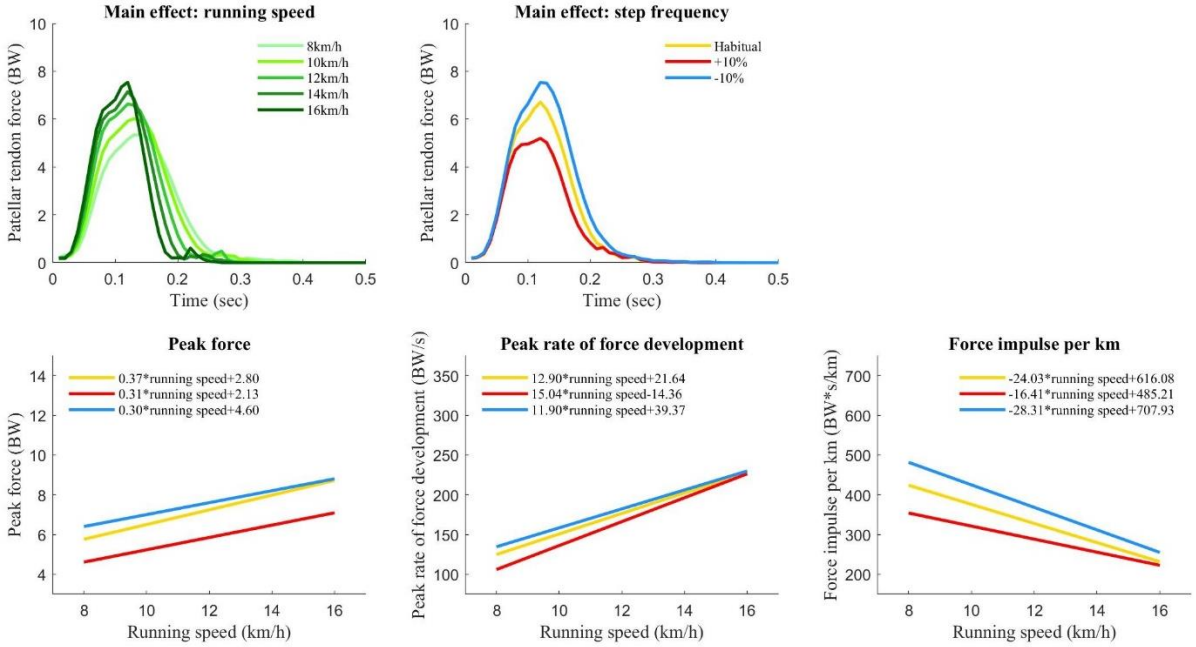

Figure 2.2. *Upper row: mean values of the PTF -expressed in bodyweight (BW)- over time. Values are grouped according to running speed, combining all step frequencies (left) and according to step frequency, combining all running speeds (right). Lower row: regression lines peak PTF (left), peak rate of PTF development (middle) and PTF impulse per kilometre (right). Red lines represent the increased step frequency condition, yellow lines the habitual step frequency condition and blue lines the decreased step frequency condition.*

Peak PTF

	Mean ± SD values (BW)		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
8km/h	6.77 ± 1.84	5.73 ± 1.48	4.58 ± 1.52
10km/h	7.77 ± 1.75	6.48 ± 1.76	5.34 ± 1.63
12km/h	8.31 ± 1.91	7.21 ± 1.71	5.87 ± 1.79
14km/h	8.85 ± 2.41	8.05 ± 1.76	6.42 ± 2.08
16km/h	9.22 ± 2.18	8.63 ± 2.14	7.16 ± 2.11
	Two-way repeated measures ANOVA		
	Running speed	**Step frequency**	**Speed*Frequency**
F-value	27.868	45.036	1.096
p-value	<0.001	<0.001	0.376
	Regression analysis		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
Coefficient	0.30	0.37	0.31
Intercept	4.60	2.80	2.13

Peak rate of PTF development

	Mean ± SD values (BW/s)		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
8km/h	130.48 ± 36.27	123.57 ± 32.59	111.74 ± 27.33
10km/h	163.50 ± 45.13	148.55 ± 35.96	144.14 ± 42.29
12km/h	183.27 ± 46.91	185.87 ± 56.08	149.23 ± 46.03
14km/h	204.72 ± 65.57	194.49 ± 53.21	182.82 ± 46.59
16km/h	228.88 ± 62.20	229.57 ± 51.61	242.82 ± 82.96
	Two-way repeated measures ANOVA		
	Running speed	**Step frequency**	**Speed*Frequency**
F-value	32.256	1.686	1.124
p-value	<0.001	0.213	0.354
	Regression analysis		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
Coefficient	11.90	12.90	15.04
Intercept	39.37	21.64	-14.36

PTF impulse per km

	Mean ± SD values (BW*s/km)		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
8km/h	488.76 ± 147.34	436.93 ± 121.91	357.49 ± 125.83
10km/h	414.78 ± 119.38	363.90 ± 108.80	315.55 ± 106.90
12km/h	370.90 ± 81.98	319.27 ± 88.05	290.49 ± 82.66
14km/h	307.47 ± 67.42	280.27 ± 65.12	253.22 ± 68.66
16km/h	259.34 ± 55.54	238.49 ± 54.17	224.52 ± 52.40
	Two-way repeated measures ANOVA		
	Running speed	**Step frequency**	**Speed*Frequency**
F-value	31.785	68.768	7.994
p-value	<0.001	<0.001	<0.001
	Regression analysis		
	Decreased frequency	**Habitual frequency**	**Increased frequency**
Coefficient	-28.31	-24.03	-16.41
Intercept	707.93	616.08	485.21

Table 2.3. *Mean values ± standard deviations and results of the two-way repeated measures ANOVA and regression analyses for PTF load parameters.*

2.4 Discussion

The objective of this study was to investigate whether the load-modifying effect of step frequency alterations was similar at different running speeds. Our hypothesis was partially confirmed: as the running speed increased, the effect of step frequency alterations on the impulse per km decreased. There was a large difference in the impulse per km between the step frequency conditions at lower running speed (8km/h), while the differences between the step frequency conditions were marginal at higher running speed (16km/h). This can be explained by the fact that increasing the step frequency does not automatically decrease the contact time with the same amount (see Supplementary analysis B). The step frequency can also be increased by reducing the flight time. Therefore, if the goal of altering the step frequency is to influence the cumulative load on the patellofemoral joint or the patellar tendon, the intervention might be more effective during slow jogging compared to high speed interval training. Future studies could even raise the question whether the effect might reverse and increasing the step frequency would increase the cumulative knee load when running speed is increased beyond 16km/h. This

is well possible since earlier research from Dorn et al. (2012) investigating the transition from medium-paced to sprinting-paced running speeds, has shown changes in the behaviour of the different parts of the quadriceps (Dorn et al., 2012). The force production of the rectus femoris during push-off and swing increased exponentially to increase the step frequency, while the force production of the vasti muscles during the breaking phase of stance reaches a steady state during sprinting because of the minimal ground contact time (Dorn et al., 2012). All this will likely have an important effect on the PFJS-time and PTF-time curves and consequentially on the PFJS and PTF impulse per km.

The effect of step frequency alteration on peak PFJS and PTF was not influenced by running speed. Peak PFJS and PTF decreased by the same amount with increasing step frequency irrespective of the running speed. Furthermore, the peak PFJS and PTF increased with increasing running speed irrespective of the step frequency. These results are in agreement with previous studies indicating that peak knee load values increase with increasing running speed (Petersen et al., 2014, 2015; Starbuck et al., 2021) and decrease when the step frequency is increased (Heiderscheit et al., 2011; Lenhart et al., 2014; Willson et al., 2014). Peak knee load values decrease at an increased step frequency because of a smaller peak knee flexion angle during stance and a smaller vertical excursion in the centre of mass, resulting in a reduced peak vertical GRF (Schubert et al., 2014). Apparently, the rate with which these parameters change is not influenced by running speed. Peak rate of PFJS and PTF development only showed a significant increase with increasing running speed and was not significantly influenced by step frequency alterations. Since the sample size was calculated to reveal differences with a medium to large effect size, is possible that a difference with a small effect size may have remained unnoticed. Nevertheless, when the intention is to alter the peak rate of PFJS and PTF development, changing the running speed would be the preferred approach. We hypothesised that an interaction effect would be present in peak PFJS and PTF and peak rate of PFJS and PTF development since the combination of a higher running speed and a higher step frequency might form an incentive for the participants to change their running style, possibly including a shift from a heel strike pattern to a midfoot-forefoot strike pattern. Vannatta et al. (2015) demonstrated that female long-distance runners who changed their habitual rearfoot strike pattern for a forefoot strike pattern at running speeds of 3.52-3.89m/s (12.67-14.00km/h) significantly reduced their peak PFJS (Vannatta & Kernozek, 2015). However, Allen et al. (2016) showed already at habitual running speed that a 10% increase in step frequency is typically not enough to automatically change from a heel strike pattern to a midfoot-forefoot strike pattern (Allen et al., 2016). Additionally, in the study of Breine et al. (2014) only 36% of

the 55 participants switched from a heel strike pattern to a midfoot-forefoot strike pattern when running speed was increased from 3.2m/s to 6.2m/s (11.52-22.32km/h) (Breine et al., 2014). Post hoc analysis of our data indicated that not a single participant altered their foot strike pattern throughout the different running speed and step frequency conditions (three of the 12 analysed participants ran consistently with a midfoot/forefoot strike pattern). Nevertheless, the interaction effect between foot strike pattern, running speed and step frequency on patellofemoral and patellar tendon load parameters can be an interesting topic for future research.

Some limitations in this study have to be reported. First, in the absence of devices that can measure PFJS or PTF directly in vivo in young, healthy individuals, musculoskeletal modelling was used to calculate these load parameters. Since a musculoskeletal model is an approximation and simplification, it comes with some assumptions and limitations. For example, not all muscle-tendon parameters of the model were scaled to subject-specific measures. This is because parameters like optimal fibre length and tendon slack length are difficult to scale accurately, even with advanced medical imaging (Blemker et al., 2007). As a consequence, the maximum isometric muscle forces were set to three times the values of the generic model to guarantee sufficient muscle force during fast concentric contractions. Furthermore, patellofemoral contact areas were determined based on generic regression equations rather than individual medical imaging. Finally, the patellofemoral kinematics were controlled by a generic spline function, possibly missing patellofemoral maltracking (Powers, 2010). However, this study has a repeated measures design. Therefore, it could be argued that an error in the calculated load parameters was similar for all conditions within a participant and the statistical analysis was not affected as such. A second limitation is the step frequency alteration throughout the running protocol. Although a metronome is frequently used for step frequency alterations in running research (Schubert et al., 2014), the method is not perfectly free from error. The predetermined percentage step frequency alteration was 10%. The fact that the average increase or decrease in step frequency for all participants was not always perfectly 10% for every running speed, probably influenced the effect size, but does not pose a threat to the main objective of the study. It is however questionable whether more familiarisation time would have reduced the variation in the step frequency. Running for one minute at each condition, and recording only the last 15 seconds, the participants had about 45 seconds to accommodate to each running condition. Whilst one may observe slowly continued adaptations in terms of running style after the initial accommodation to a condition, these would be expected to be very subtle. Additionally, more time at the higher running speeds could potentially generate more

variation because of increased fatigue. A final limitation of this study is the absence of patients with patellofemoral pain syndrome or patellar tendinopathy. This study has included recreational runners, who have a higher risk of developing a running-related injury (Nielsen et al., 2012; van Poppel et al., 2018). However, the recreational runners in this study were free of injury for at least six months. In patients with patellofemoral pain syndrome or patellar tendinopathy, a maladaptive running pattern might be the cause of the running-related injury. Conversely, pain might lead patients towards an altered running pattern that favours decreased knee loads. In these scenarios, this study might either under- or overestimate the effect of alterations in running speed or step frequency on PFJS and PTF parameters. However, previous studies did not find significant differences in PFJS between patients with patellofemoral pain syndrome and healthy individuals during running (Willson et al., 2014; Wirtz et al., 2012).

Even in the absence of a patient group, this study has clinical implications. The results of this study can be used to develop a symptom modification test (Schubert et al., 2014). First, the test could decide if maladaptive loading is a dominant factor for the development of patellofemoral pain syndrome or patellar tendinopathy. That is, if a patient can run a longer distance without pain in the patellofemoral joint or the patellar tendon if the running speed or the step frequency is altered, then load management may be considered as part of the rehabilitation. If the pain is not influenced by these alterations, other contributing factors and interventions need to be considered. Second, if maladaptive loading is a dominant factor and load management is considered as intervention, the test can help identify which load parameter should be controlled. Namely, if a patient gets pain while running a certain distance at low speed, but not while running that same distance at high speed, the cumulative load is probably the dominant factor that causes the pain. This hypothesis can be confirmed when the pain disappears after an increase in step frequency at a low running speed. However, if a patient gets pain while running a certain distance at high speed but not while running that same distance at low speed, the peak load or peak rate of load development is probably the dominant factor that causes the pain. Since an increased step frequency at high speed significantly decreases the peak load, but not the peak rate of load development, step frequency alteration can be used to differentiate between both loading parameters as a dominant cause of pain. The exact running distance and running speed used during this symptom modification test, depend on the history of the individual patient in terms of habitual running parameters and the running distance and speed at which the pain is provoked. Furthermore, some patients may need a larger alteration in running speed and step frequency than others before the intervention becomes effective. Future studies with actual patients should confirm the effectiveness of this symptom modification test.

2.5 Conclusion

The load-reducing effect of increasing the step frequency on the impulse per km diminishes at faster running speed. Peak PFJS and PTF or peak rate of PFJS and PTF development per step increased with increasing running speed irrespective of the step frequency and the peak PFJS and PTF was reduced by increasing the step frequency, irrespective of the running speed. Clinicians can use this knowledge in their clinical examination to develop load management strategies for the treatment of patellofemoral pain syndrome and patellar tendinopathy.

2.6 References

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Supplementary analysis A

	Decreased frequency	Habitual frequency	Increased frequency
8km/h	5 ± 7 (0 ± 3)	5 ± 7	5 ± 6 (0 ± 2)
10km/h	9 ± 10 (-1 ± 4)	10 ± 11	11 ± 9 (1 ± 5)
12km/h	21 ± 17 (1 ± 8)	20 ± 15	22 ± 13 (2 ± 9)
14km/h	34 ± 21 (-2 ± 16)	36 ± 20	38 ± 17 (2 ± 14)
16km/h	46 ± 22 (-6 ± 12)	52 ± 21	53 ± 20 (1 ± 15)

Table 2.4. *Mean values ± standard deviations for rate of perceived leg muscle exertion. The top values represent absolute values on the CR100 scale. The bottom values between brackets in the decreased and increased frequency condition represent relative percentages compared to the habitual frequency condition at the same running speed.*

Supplementary analysis B

Contact time (s)

	Decreased frequency	Habitual frequency	Increased frequency
8km/h	0.33 ± 0.04	0.31 ± 0.03	0.29 ± 0.03
10km/h	0.29 ± 0.02	0.28 ± 0.02	0.27 ± 0.03
12km/h	0.26 ± 0.02	0.25 ± 0.02	0.24 ± 0.02
14km/h	0.24 ± 0.02	0.24 ± 0.02	0.23 ± 0.02
16km/h	0.22 ± 0.01	0.22 ± 0.01	0.21 ± 0.02

Stride length (m)

	Decreased frequency	Habitual frequency	Increased frequency
8km/h	0.33 ± 0.04	0.31 ± 0.03	0.29 ± 0.03
10km/h	0.29 ± 0.02	0.28 ± 0.02	0.27 ± 0.03
12km/h	0.26 ± 0.02	0.25 ± 0.02	0.24 ± 0.02
14km/h	0.24 ± 0.02	0.24 ± 0.02	0.23 ± 0.02
16km/h	0.22 ± 0.01	0.22 ± 0.01	0.21 ± 0.02

Strides per km

	Decreased frequency	Habitual frequency	Increased frequency
8km/h	528 ± 63	595 ± 27	655 ± 37
10km/h	431 ± 46	480 ± 23	533 ± 28
12km/h	409 ± 68	418 ± 23	485 ± 86
14km/h	364 ± 55	374 ± 23	429 ± 68
16km/h	330 ± 47	337 ± 22	383 ± 61

Table 2.5. *Mean values ± standard deviations for spatial-temporal parameters during the different running conditions.*

Chapter 3

Study 2: External weight mass and carrying position influence peak patellar tendon force and patellofemoral joint contact force independently during forward lunge

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Abstract

Objectives

The forward lunge is a common exercise in the rehabilitation of patellar tendinopathy and patellofemoral pain syndrome. External weights are frequently used to increase the peak patellar tendon force and patellofemoral joint contact force during this exercise. The weight's position might influence this relationship. The objective of this study was to investigate the combined effect of an external weight's mass and carrying position on the peak patellar tendon force and patellofemoral joint contact force during a forward lunge.

Methods

Ten healthy individuals performed forward lunges holding external weights between 0.1 to 0.3 times body mass either in one hand at the ipsilateral or contralateral side of the leading leg, or in two hands at the side or in front of the trunk. Three-dimensional kinematic data and ground reaction forces were collected and peak patellar tendon force and patellofemoral joint contact force were calculated using musculoskeletal modelling. Two-way repeated measures ANOVA's determined the main effects for the external weight's mass and position as well as their interaction effect.

Results

Increasing the mass of the external weights increased both the peak patellar tendon force and patellofemoral joint contact force linearly and at the same rate in all positions. Both peak forces were larger in the one-hand ipsilateral and two-hand side positions.

Conclusion

An external weight's mass and position both influence the peak patellar tendon force and patellofemoral joint contact force during a forward lunge. The rate of increase in peak forces with increasing mass was similar for all weight-carrying positions.

Key words

Exercise therapy; patellar tendon; patellofemoral joint

3.1 Introduction

Patellar tendinopathy (PT) and patellofemoral pain syndrome (PFPS) are two common and disabling injuries among the young, physically active population (Francis et al., 2019; Lian et al., 2005; Smith et al., 2018; Vries et al., 2017; Johannes Zwerver et al., 2011). Both pathologies are considered overuse injuries (Cook & Purdam, 2009; Dye, 2005). Therefore, searching a balance between the load imposed on the painful structure and load capacity of that structure is an important aspect in the rehabilitation of PT and PFPS (Lack et al., 2018; Malliaras et al., 2015). Exercise programs for patients with PT or PFPS are based on the principle of mechanotherapy: stimulating the adaptive capacity of the injured structures through an optimal and progressively increasing dose of mechanical load (Khan & Scott, 2009). Important load parameters in this context are the peak loads on the patellar tendon or the patellofemoral joint (Dye, 2005; Malliaras et al., 2015). Next to load, it is increasingly recognised that muscle weaknesses and impaired control of joint motions throughout the kinetic chain of the lower limb and trunk are involved in causing these knee injuries (Lack et al., 2018; Malliaras et al., 2015; Powers, 2010; Silva et al., 2015). In consequence, strengthening and improving neuromuscular control of muscles in this kinetic chain through functional, multi-joint exercises is an integral part of the exercise programs for PT and PFPS. Therefore, the forward lunge is an exercise that can be used as part of an exercise program with progressive loading of the patellar tendon and patellofemoral joint. This exercise loads the knee as part of a functional movement involving multiple joints and several muscle groups.

External weights in the form of barbells, dumbbells or kettlebells are commonly used to progressively increase the peak patellar tendon force (PTF) and the patellofemoral joint contact force (PFJCF) during the forward lunge. Riemann et al. (2012) reported that the net knee extensor moment impulse increased linearly when the mass of the external weights was increased up to 0.5 times body mass (BM). Furthermore, Stastny et al. (2015) found that vastus lateralis activity in the leading leg was higher when the same external weight was held in the contralateral hand compared to the ipsilateral hand. However, no prior study has investigated the effect of either the mass or the position of an external weight on peak PTF or PFJCF specifically. Furthermore, the combined effect of mass and carrying position on peak PTF and PFJCF has not yet been investigated. It is plausible that an increase in mass of the external weight does not increase the peak PTF and PFJCF with the same amount in each weight-carrying position. This is because the weight-carrying position does not only change joint moments and muscle activations at the level of the knee but also the hip (Riemann et al., 2012;

Stastny et al., 2015). It might therefore be possible, for example, that combined changes in the mass and carrying position of the external weight alters co-contractions between the quadriceps and hamstrings, which could have consequences for the peak PTF and PFJCF. A method that can verify this hypothesis, is musculoskeletal modelling using static optimisation to calculate the peak PTF and joint reaction analysis for the peak PFJCF (Delp et al., 1990; Goulette et al., 2021; Zellmer et al., 2019). This method can determine our load parameters of interest while accounting for the influence of other joint moments and muscle forces within the kinetic chain. The objective of this study was to investigate the combined effects of an external weight's mass and carrying position on the peak PTF and PFJCF during a forward lunge. We hypothesised that peak PTF and PFJCF will increase linearly with increasing mass of the external weight. We also hypothesised that differences in the peak PTF and PFJCF exist with different weight-carrying positions, but that these differences will alter as the mass of the external weight increases.

3.2. Methods

3.2.1. Participants

Ten healthy individuals (five males and five females) participated in this study. The participants were free from any lower limb injury or low back pain in the past six months and had no history of major lower limb injury that required surgery. The participants were 24 ± 3 years old, had a BM of 69 ± 14 kg and a body height of 174 ± 10 cm (mean ± standard deviation). The study was approved by the ethical committee of UZ/KU Leuven, Belgium (reference number: S62706). All participants provided written informed consent before the start of data collection. The minimum required sample size was calculated in G*Power 3.1 (Erdfelder et al., 2009) before the start of the data collection. For an arbitrary moderate effect size Cohen's f of 0.3 and an alpha of 0.025, it would take six participants to detect significant effects with a statistical power of 0.80 or higher.

3.2.2. Protocol

The forward lunges were performed with the dominant leg as the leading leg. The participants started in a standing position with both feet next to each other. Next, the participants stepped forward with the leading leg and lowered the body until the thigh of the leading leg was horizontal. The step length was 1.0 times the leading leg length -measured as the distance from the anterior superior iliac spine to the medial malleolus- and was marked with tape on the floor. Finally, the participants pushed themselves back up towards the standing position with both feet together. No further instructions or constraints were given regarding movement execution or timing to make the results of this study widely generalisable.

External weights were equal to 0.1, 0.2 and 0.3 BM. These external weights were held in four different positions: one weight in both hands in front of the trunk (2HF), one weight in each hand at both sides of the trunk (2HS), one weight in one hand at the side of the trunk ipsilateral of the leading leg (1HIL) and one weight in one hand at the side of the trunk contralateral of the leading leg (1HCL) (see Figure 3.1). The participants performed four repetitions of each variation.

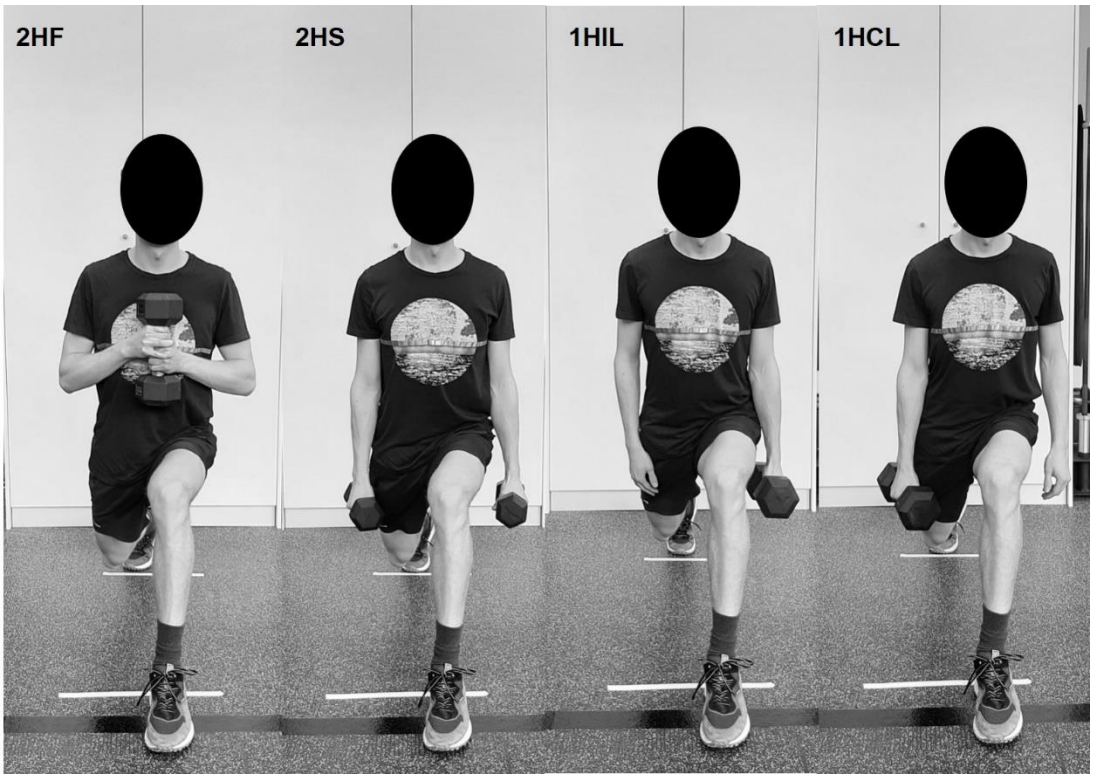

Figure 3.1. *Weight-carrying positions*

2HF = two hands in front of the trunk; 2HS = two hands aside the trunk; 1HIL = one hand ipsilateral of leading leg; 1HCL = one hand contralateral of leading leg.

3.2.3. Data collection

Three-dimensional kinematic data were collected with a passive marker-based motion capture system (Vicon, Oxford, UK). Ten MX-T20 cameras measured marker positions at a sampling rate of 100Hz. Fifty three anatomical markers and five rigid clusters of three makers were placed on the legs, trunk and arms according to an extended full-body Plug-In Gait model (Davis et al., 1991). The ground reaction force (GRF) under the leading leg was measured at a sampling frequency of 1000 Hz using a floor-embedded force plate (type OR6-7, AMTI, Watertown, MA, USA).

3.2.4. Data processing

Only the part of the forward lunge where the leading leg was in contact with the force plate, was retained for further analysis. A 50 N threshold of the vertical component of the GRF was used to mark initial contact and push-off.

PTF and PFJCF were calculated in OpenSim 3.3 (OpenSim, Stanford, CA, USA) (Delp et al., 2007), using the Catelli model as generic musculoskeletal model, which is customised for deep hip and knee flexion tasks (Catelli et al., 2019). The lower limbs of this model consisted of a 6 DoF pelvis, a 3 DoF hip joint, a 1 DoF tibiofemoral joint, a 1 DoF patellofemoral joint, and a 1 DoF ankle joint. The motions of the lower limb were driven by 80 Hill-type muscle-tendon units of which the properties are based on the MRI data of young, physically active individuals (Rajagopal et al., 2016). Motions of the trunk and arm segments were driven by ideal torque actuators. The spatial dimensions of the segments, the optimal muscle fibre lengths and tendon slack lengths in the generic model were scaled to the values of the individual participants using the marker data from a static calibration trial. The maximum isometric muscle forces were scaled to the participant's BM and body height (Handsfield et al., 2014). The kinematics of the model were determined using a global optimisation approach, minimising the weighted sum of squared differences between the measured marker positions and the modelled markers (Lu & O'Connor, 1999). The movement of the patella was not controlled by markers, but by a spline function with the tibiofemoral flexion angle as a determinative variable. Next, net joint moments were calculated using inverse dynamics based on the kinematics results and the GRF data, both low-pass filtered with a cut-off frequency of 6Hz (Bisseling & Hof, 2006). Forces generated by individual muscle-tendon units to balance the net joint moments, were calculated with a static optimisation approach (Mokhtarzadeh et al., 2014), in which the sum of all muscle activations squared was minimized. The PTF was calculated as the scalar sum of forces produced by the four quadriceps muscle-tendon units (Hagen, Vanmechelen, et al., 2023). It

was possible to take the scalar sum since the four muscle-tendon units of the quadriceps run in parallel at the level of the patellar tendon. Finally, the kinematics, GRF, and muscle-tendon forces data were used to calculate joint reaction forces in the OpenSim joint reaction analysis tool (Steele et al., 2012). The PFJCF was defined as the patellofemoral joint reaction force perpendicular to the patellofemoral joint surface. Both peak PTF and peak PFJCF were expressed in relation to the participants' body weight (BW) i.e. the participants' BM multiplied by 9.81m/s^2. Peak PTF and PFJCF values were identified over the entire trial. The average peak PTF and PFJCF values of the four trials per participant per condition were used for the statistical analysis.

3.2.5. Statistical analysis

The statistical analysis was performed in the open-source software package SPM1D M0.4.9 (https://spm1d.org) (Pataky, 2012). Peak PTF and PFJCF were analysed with zero-dimensional two-way repeated measures ANOVA's (mass * position). Since the data of both parameters were not normally distributed, non-parametric ANOVA's based on permutation tests were chosen (Nichols & Holmes, 2003). The level of significance was set to 0.025, applying a Bonferroni correction by dividing a standard alpha value of 0.05 by the number of load parameters, i.e. 2. In case of a significant main effect for the weight-carrying position, permutation-based paired t-tests were used as post-hoc analysis to identify differences between the positions. In case of a significant main effect for the mass of the external weight or a significant interaction effect, a regression analysis was chosen as post-hoc analysis. After confirming linear trends through visual inspection, a separate linear regression equation was calculated for each weight-carrying position. The mass of the external weight was the independent variable and the peak PTF or PFJCF was the dependent variable. The regression coefficients represent the effect size of the main effect for the mass of the external weight, since they tell how much the peak PTF or PFJCF would increase for every 1 BW increase in external weight. In case of a significant main effect for the mass of the external weight, these regression coefficients would be significantly different from zero, representing a horizontal line. In case of a significant interaction effect, the regression lines would not be parallel and the regression coefficients would be significantly different from one another.

3.3. Results

3.3.1. Patellar tendon force

Mean values and standard deviations for the different lunge conditions and the results of the two-way repeated measures ANOVA for the peak PTF can be found in Table 3.1.A. Significant main effects for the mass ($p<0.001$) and position ($p=0.001$) of the external weights were found, but no significant interaction effect ($p=0.290$) was found. Post-hoc analyses for the weight-carrying position showed that the peak PTF for the 1HCL and the 2HF conditions were significantly lower than the 2HS and 1HIL conditions (see Table 3.1.B and Figure 3.2). In the regression analysis, the regression intercepts showed clear differences between the 1HCL and the 2HF conditions on one hand and the 2HS and 1HIL conditions on the other hand. The regression coefficients for the mass of the external weight were 4.14 for the 1HCL condition, 6.24 for the 2HF condition, 6.25 for the 2HS condition and 8.87 for the 1HIL condition. The fact that there is no significant interaction effect can be seen by the regression lines that run rather parallel.

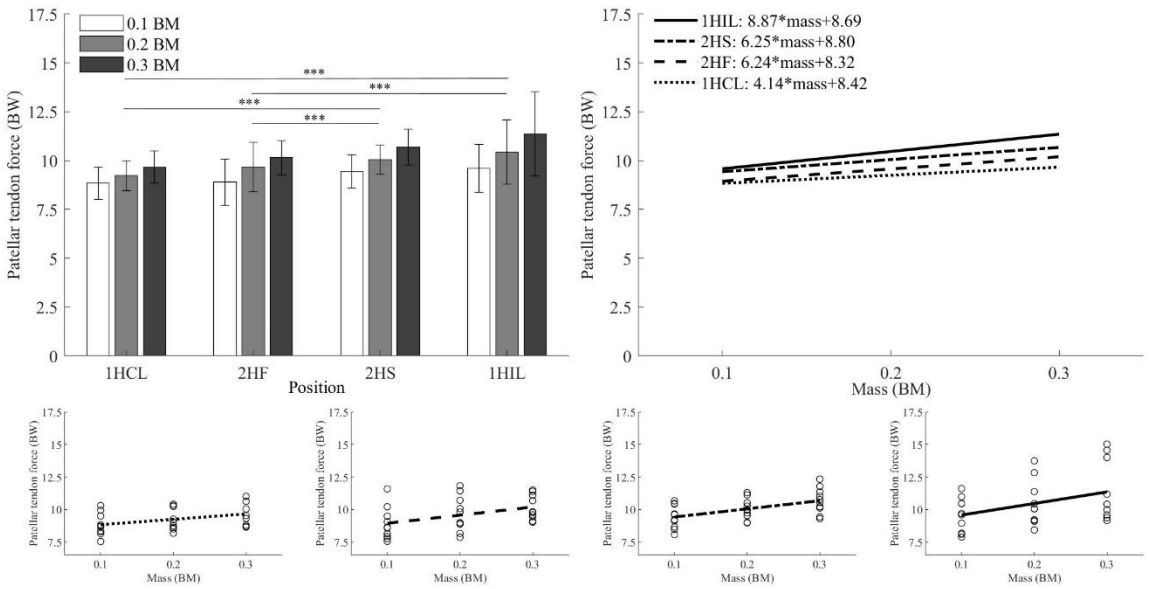

Figure 3.2. *Mean values ± standard deviations and linear regression analyses for peak patellar tendon force.*
**** = p<0.001*

	Mean ± SD values (BW)		
	+0.1 BM	**+0.2 BM**	**+0.3 BM**
2HF	8.90 ± 1.19	9.66 ± 1.27	10.15 ± 0.89
2HS	9.44 ± 0.85	10.04 ± 0.75	10.69 ± 0.92
1HIL	9.59 ± 1.23	10.44 ± 1.64	11.37 ± 2.15
1HCL	8.83 ± 0.81	9.23 ± 0.76	9.66 ± 0.83
	Two-way repeated measures ANOVA		
	Mass	**Position**	**Mass*Position**
F-value	34.415	7.287	1.096
p-value	<0.001	0.001	0.290
Partial eta squared	0.368	0.157	0.100

Table 3.1.A. *Mean values ± standard deviations and results of the two-way repeated measures ANOVA for peak patellar tendon force.*

	2HS	**1HIL**	**1HCL**
2HF	t=3.255 p<0.001 d=1.029	t=3.602 p<0.001 d=1.139	t=1.818 p=0.069 d=0.575
2HS	x	t=1.984 p=0.057 d=0.627	t=6.692 p<0.001 d=2.116
1HIL	x	x	t=4.812 p<0.001 d=1.522

Table 3.1.B. *T-scores, p-values and Cohen's d effect sizes for the differences between the weight-carrying positions for peak patellar tendon force.*

3.3.2. Patellofemoral joint contact force

Descriptive values and the results of the two-way repeated measures ANOVA for the peak PFJCF can be found in Table 3.2.A. The results for peak PFJCF were similar to the results of the peak PTF with a significant main effect for the mass ($p<0.001$) and position ($p<0.001$) of the external weights without a significant interaction effect ($p=0.206$). Post-hoc t-tests also showed significantly lower peak PFJCF values in the 1HCL and the 2HF conditions compared to the 2HS and 1HIL conditions (see Table 3.2.B and Figure 3.3). The regression coefficients for the mass of the external weight altered between 4.65 in the 1HCL condition and 10.24 in the 1HIL condition. The regression lines were parallel for the peak PFJCF as well.

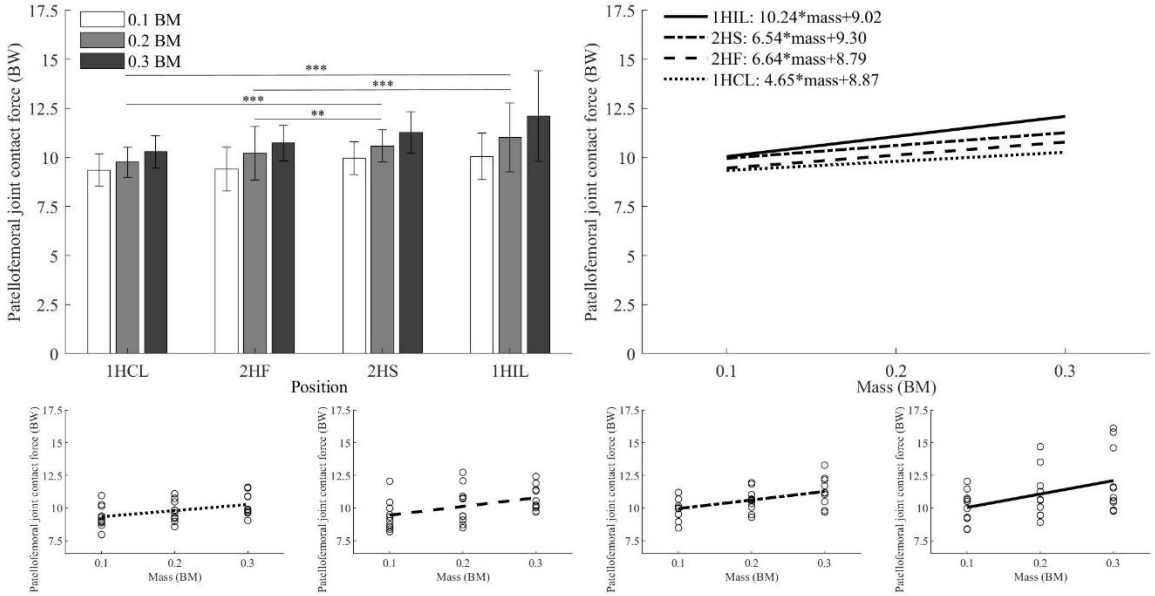

Figure 3.3. *Mean values ± standard deviations and linear regression analyses for peak patellofemoral joint contact force.*

** = $p<0.01$

*** = $p<0.001$

	Mean ± SD values (BW)		
	+0.1 BM	**+0.2 BM**	**+0.3 BM**
2HF	9.41 ± 1.11	10.22 ± 1.37	10.74 ± 0.90
2HS	9.96 ± 0.85	10.59 ± 0.83	11.27 ± 1.07
1HIL	10.06 ± 1.18	11.02 ± 1.76	12.11 ± 2.32
1HCL	9.36 ± 0.83	9.76 ± 0.76	10.29 ± 0.82
	Two-way repeated measures ANOVA		
	Mass	**Position**	**Mass*Position**
F-value	31.443	7.223	1.463
p-value	<0.001	<0.001	0.206
Partial eta squared	0.348	0.156	0.129

Table 3.2.A. *Mean values ± standard deviations and results of the two-way repeated measures ANOVA for peak patellofemoral joint contact force.*

	2HS	**1HIL**	**1HCL**
2HF	t=3.055 p=0.002 d=0.966	t=3.585 p<0.001 d=1.134	t=1.724 p=0.085 d=0.545
2HS	x	t=1.971 p=0.057 d=0.623	t=6.313 p<0.001 d=1.996
1HIL	x	x	t=4.661 p<0.001 d=1.474

Table 3.2.B. *T-scores, p-values and Cohen's d effect sizes for the differences between the weight-carrying positions for peak patellofemoral joint contact force.*

3.4. Discussion

Exercise programs for patients with PT or PFPS are designed to increase the load capacity of the injured structures through an optimal and progressively increasing dose of mechanical load (Lack et al., 2018; Malliaras et al., 2015). In this context it is important to know how patellar tendon and patellofemoral load parameters respond to variations within rehabilitation exercises. The objective of this particular study was to investigate the influence of the mass and the position of an external weight on the peak PTF and PFJCF during a forward lunge. As expected, peak PTF and PFJCF increased significantly with increasing mass of the external weight. More importantly, peak PTF and PFJCF showed linear increases while the mass was raised from 0.1 to 0.3 BM. For every 0.1 BM increase in mass of the external weights, the peak PTF increased with 0.414 to 0.887 BW and peak PFJCF increased with 0.465 to 1.024 BW. Taking the correction factor of 9.81 m/s^2 into account for the conversion from body mass to body weight, we can conclude that the peak PTF and PFJCF increase at a slower rate than the mass of the external weight. Similar to this study, Riemann et al. (2012) found a linear increase in the net knee extensor moment impulse while the mass of the external weight was raised from 0 to 0.5 BM, holding external weights in the 2HS position. However, research in heavy load squatting has shown a curvilinear relationship in the peak knee extensor moment, the knee extensor work and the quadriceps muscle activity which did not further increase after 0.8 times the one repetition maximum despite an increase in the mass of the external weight (Armstrong et al., 2022; Flanagan & Salem, 2008; van den Tillaar et al., 2019). It would be an interesting topic of future research to determine if and when the peak PTF and PFJCF start to show a similar plateauing effect during the forward lunge.

In order to progressively increase the peak PTF and PFJCF during a forward lunge, it is not always necessary to increase the mass of the external weight. The results show that peak PTF and PFJCF values were also significantly influenced by the weight-carrying position. Peak PTF and PFJCF were significantly larger in the 1HIL and 2HS position compared to the 1HCL and 2HF position. Post-hoc analysis of the GRF under the leading leg provides evidence that this main position effect can be predominantly explained by the effect of the weight-carrying position on the point of application of the GRF. In the frontal plane, changing the external weight from the contralateral side to the ipsilateral side of the leading leg increases the portion of the GRF applied to the leading leg compared to the trailing leg. In the sagittal plane, the combined centre of mass of the participant and the external weight shifts towards the knee of the leading leg when the weight is placed in front of the trunk compared to at the side of the

trunk. This shift leads to a decrease in the moment arm of the GRF compared to the knee joint. Another possible explanation for the change in peak PTF and PFJCF between weight-carrying positions, could be changes in the movement execution and kinematics between the conditions. However, even though instructions and constraints regarding movement execution were minimal, key kinematic parameters remained stable over the different weight-carrying positions. For the mean values of the peak knee flexion angle, the maximum difference between weight-carrying positions was 6°. Similar conclusions can be drawn at the level of the trunk. The maximum difference between the mean values of two weight-carrying positions was <1° for the peak trunk lateral flexion angle and 7° for the peak trunk flexion angle.

Changing the external weight's mass and carrying position influences not only the knee, but also the hip and ankle regions. Because of co-contractions from bi-articular muscles such as the hamstrings and gastrocnemius, we cannot automatically assume that an increase in the mass of the external weights increases the peak PTF and PFJCF with the same rate in each weight-carrying position. In other words, there may be an interaction effect of the external weight's mass and carrying position on the peak PTF and PFJCF because of muscular co-contractions. The results show however that no such interaction effect was present and that the peak PTF and PFJCF increased at the same rate over the different weight carrying positions.

This study comes with some limitations. First, the participants in this study were all free from injury. The question could be raised if patients with PT or PFPS would show a different movement pattern because of pain apprehension. However, if such difference in movement pattern between patients and healthy individuals would exist, it is still questionable whether alterations an external weights' mass or the carrying position would have a different effect in both groups. A second limitation of this study are the assumptions and simplifications made during the musculoskeletal modelling process. For example, the movement of the patella is controlled by a generic spline function (Catelli et al., 2019; Rajagopal et al., 2016). Patellofemoral malalignment or maltracking can therefore not be included in the calculations of the peak PTF or PFJCF. This is because the position and movement of the patellofemoral joint can only be accurately measured with advanced medical imaging techniques (Wheatley et al., 2020; Yu et al., 2019). Another example, is the assumption of a fixed patellar tendon length during the static optimisation calculations. On the other hand, because of the repeated measures design of this study, an offset in the peak PTF or PFJCF was expected to be similar for all conditions within a participant and the statistical analysis was not affected as such. However, different assumptions and simplifications between models and load calculations may lead to different peak PTF and PFJCF values for the same data (Garcia et al., 2022; Kernozek et al.,

2015; Mokhtarzadeh et al., 2014). We believe that the model and analysis techniques used in this study were appropriate to answer the specific research question. However, it would not be appropriate to compare the absolute values of this study against the results of other studies that used different analysis techniques.

Of course, changing the mass and carrying position of an external weight is not the only way to manipulate the peak PTF and PFJCF during a forward lunge. Other cross-sectional studies with a controlled experimental design evaluated the effect of other parameters such as step length (Escamilla et al., 2008; Riemann et al., 2013), trunk inclination (Farrokhi et al., 2008) or foot position (Zellmer et al., 2019). These studies provide useful information to physiotherapists on how to manipulate knee loading in a general population. However, the ideal situation would be that physiotherapists can determine peak PTF and PFJCF for individual patients and exercises in real-time in clinical practice (Verheul et al., 2020). The results of this and other studies are an important step towards the development of such clinical measurement tools.

3.5. Conclusion

Physiotherapists require accurate information on load progressions during exercise variations. The results of this study can be used in clinical practice to manipulate peak PTF and PFJCF during a forward lunge. Increasing the mass of an external weight increases peak PTF and PFJCF linearly. This increase in peak PTF and PFJCF happens at a slower rate than the increase in mass. Furthermore, the peak PTF and PFJCF are larger when the external weight is carried in one hand at the ipsilateral side of the leading leg compared to the contralateral side or when the external weight is carried in two hands aside the trunk compared to in front of the trunk. Increasing the mass of the external weights increases the peak PTF and PFJCF with the same amount in the different weight carrying positions.

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Chapter 4

Study 3: Peak patellar tendon force progressions during heavy load single-leg squats on level ground and decline board

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Abstract

Objectives

Progressive tendon loading programs for patellar tendinopathy typically include single-leg squats with heavy weights either on level ground or on a decline board. Changes in patellar tendon force due to variations of the heavy load single-leg squat have not yet been objectively quantified. The objective of this study was to investigate the influence of the mass of an external weight and the use of a decline board on the peak patellar tendon force during a heavy load single-leg squat.

Methods

Twelve healthy participants performed single-leg back squats on a decline board and level ground at 70%, 80% and 90% of their one repetition maximum. Three-dimensional kinematics and ground reaction forces were measured and the peak patellar tendon force was calculated using musculoskeletal modelling. A two-way repeated measures ANOVA determined the main effects for the mass of the external weights and the use of a decline board as well as their interaction effect.

Results

Peak patellar tendon forces were significantly higher on the decline board compared to level ground ($p<0.05$). Neither on the decline board, nor on level ground did the peak patellar tendon force increase significantly when increasing the external weights ($p>0.05$).

Conclusion

Progression in peak patellar tendon forces during a heavy load single-leg squat can only be obtained with a decline board. Increasing the mass of the external weight from 70% to 90% of the one repetition maximum does not result in a progressively higher peak patellar tendon force.

Key words

Exercise therapy; patellar tendon force; single-leg squat

4.1. Introduction

Patellar tendinopathy (PT) is a common overuse injury in athletes involved in sprinting or jumping sports (Lian et al., 2005). A recent systematic review reported an overall prevalence 18.3% in the general population, increasing to 20.8% and 24.8% in basketball and volleyball players specifically (Nutarelli et al., 2023). The key symptom of PT is persistent patellar tendon pain and loss of function related to mechanical loading (Scott et al., 2020). This overuse injury is caused by an imbalance between the load a movement imposes on the patellar tendon and an individual's patellar tendon load capacity, resulting in pain, structural damage and/or loss of function (Cook et al., 2016; Wang, 2006). One form of load the patellar tendon should be able to sustain without excessive tendon strain, is the high peak patellar tendon force (PTF) during jumping (Tayfur et al., 2022; Vermeulen et al., 2023). Therefore, resistance training with heavy weights is considered a first-line treatment in both the rehabilitation and prevention of patellar tendinopathy (Breda et al., 2021; Burton, 2022; Kongsgaard et al., 2009; Lazarczuk et al., 2022; Malliaras et al., 2015).

A resistance exercise typically used to generate a high peak PTF, is the single-leg squat on a decline board (Escriche-Escuder et al., 2020). Several studies have provided evidence that peak PTF values are higher on a decline board compared to level ground (Frohm et al., 2007; Kongsgaard et al., 2006; Richards et al., 2016; J. Zwerver et al., 2007). These studies measured PTF values for the single-leg decline squat with body mass alone or with an external weight of 10kg since these variations can be performed at home with a minimum of additional equipment. Nevertheless, clinical guidelines for heavy load resistance training in the rehabilitation of patellar tendinopathy prescribe external weight masses that progressively increase between 70% and 90% of the one repetition maximum (1RM) (Escriche-Escuder et al., 2020; Malliaras et al., 2015). Unfortunately, the peak PTF has never been objectively quantified in such a heavy load situation, with the mass of the external weight being quantified as a percentage of the 1RM. Let alone that the effect of progressively increasing the mass of the external weights during the single-leg decline squat has been investigated.

On the other hand, multiple studies have already investigated the effect of adding external weights with increasing mass during a double-leg squat on level ground. Knee load parameters typically investigated are knee extensor moments and muscle activation levels from different parts of the quadriceps. These parameters do not increase linearly as the mass of the external weight approaches the 1RM, but reach a plateau (Armstrong et al., 2022; Farris et al., 2016; Flanagan & Salem, 2008; McBride et al., 2010; van den Tillaar et al., 2019). However, the

behaviour of the peak PTF during heavy double-leg squatting with increasing mass has not yet been investigated, let alone during a more unstable single-leg squat. Furthermore, it should be pointed out that the knee extensor moment is not the only factor that influences the PTF. Other influencing factors include the angle-specific moment arm of the patellar tendon, the angle-specific direction of the PTF vector and antagonistic co-contractions. These factors can be accounted for by calculating the peak PTF through advanced musculoskeletal modelling. Knowing whether or not the peak PTF reaches a plateau during single-leg squats on level ground or a decline board, would be an important piece of information for physiotherapists and strength and conditioning coaches.

The objective of this study was to investigate the influence of the mass of an external weight and the use of a decline board on the peak PTF during a heavy load single-leg squat. Based on previous research in double-leg squats (Armstrong et al., 2022; Farris et al., 2016; Flanagan & Salem, 2008; van den Tillaar et al., 2019), we hypothesised to see no significant increase in the peak PTF as the mass of external weights increases, which may be counterintuitive. Furthermore, we hypothesised that also in heavy load conditions, the decline board would increase the peak PTF.

4.2. Methods

4.2.1. Participants

Twelve individuals with at least one year of experience in strength training with heavy weights were included in this study. The required sample size was calculated in G*Power 3.1 (Erdfelder et al., 2009). For an arbitrary moderate effect size Cohen's f of 0.3 and an alpha of 0.05, 12 participants were necessary to detect significant effects with a statistical power of 0.80 or higher. Participants were recruited in local fitness centres. All participants were free from any lower limb injury or low back pain in the past six months and had no history of major lower limb injury that required surgery. Furthermore, all participants performed at least two strength training sessions per week. Participants were excluded if they had interrupted their strength training regime for more than 14 days before the start of the data collection. The study was approved by the ethical committee of UZ/KU Leuven, Belgium (reference number: S62706). All participants provided written informed consent before the start of data collection.

4.2.2. Procedure

The participants performed a single-leg barbell back squat up to a knee flexion angle of 90° on two surfaces: level ground and a 20° decline board (see Figure 4.1) (Richards et al., 2016; J. Zwerver et al., 2007). A standard 20 kg Olympic male barbell and weighted plates were used as external load. To avoid a potential systematic effect of limb dominance, half of the participants performed the single-leg squats with their dominant leg, while the other half performed the squats with their non-dominant leg.



Figure 4.1. *Heavy load single-leg squat to 90° knee flexion on a decline board (left) and level ground (right).*

The data collection sessions were spread over two testing days. On the first day, the 1RM was determined for each surface. The participants started with a self-selected warm-up. Next, one test leader explained and demonstrated the single-leg squat on both level ground and on the decline board, and the participants performed familiarisation trials with only the barbell as external weight. During these familiarisation trials, the squat depth to attain a knee flexion angle of 90° was determined for each surface. The knee flexion angle was measured manually with a goniometer. A chair with adjustable height was used to standardise the squat depth. The actual 1RM determination was always performed on the decline board first, followed by the level ground condition since we hypothesised that the 1RM on the decline board would be lower. Each participant performed one set of ten repetitions at 50% of the estimated 1RM and one set of five repetitions at 80% of the estimated 1RM. This estimated 1RM was calculated as 50% of the participant's 1RM for the double leg squat on level ground. Next, the participants continued with sets of one repetition starting at 95% of the estimated 1RM and increasing the weight with 5 kg after each successful attempt. The rest period between attempts was 3 minutes, but could be extended at the participant's request. An attempt was considered successful if the participant was able to slowly descent, gently touch the chair with their glutes and push themselves back up without losing balance or bringing the other foot down to the floor. The 1RM was the largest weight that could be lifted successfully one time. If three failed attempts were registered, the weight was considered too heavy.

On the second day, biomechanical data were collected. This biomechanical data collection session took place two weeks after the 1RM session. The participants started with a warm-up consisting of 5 minutes of cycling on a stationary bike at self-selected speed and resistance and a self-selected amount of single-leg squats with increasing weight until the participants felt comfortable to start with the recorded trials. Next, five successful squat trials were recorded, both on level ground and on the decline board, for 70%, 80% and 90% of their respective 1RM. To ensure standardised squat depth, the chair was placed at the same distance and height as in the 1RM session. All participants wore standardised footwear with flat soles and no cushioning (Indoor Copa, Kelme, Elche Spain). The order of the different conditions was block-randomized using an online randomization tool (www.randomizer.org). Trials on the same surface were always performed consecutively. In between conditions, a rest period of minimum 3 minutes was provided.

4.2.3. Biomechanical data collection

Biomechanical data collection included three-dimensional kinematic data and ground reaction forces (GRF). Kinematic data from the lower limbs and trunk (Malfait et al., 2014) were collected with 13 MX-T20 cameras of a passive marker-based system (Vicon, Oxford, UK) at a sampling rate of 100Hz. GRF data was collected with a ground-embedded force plate (type OR6-7, AMTI, Watertown, MA, USA) at a sampling frequency of 1000Hz.

4.2.4. Data processing

Musculoskeletal modelling was performed in OpenSim 3.3 (OpenSim, Stanford, CA, USA) (Delp et al., 2007). The generic model was customized for deep hip and knee flexion ranges of squatting (Catelli et al., 2019). The lower limbs of this model consisted of a 6 DoF pelvis, a 3 DoF hip joint, a 1 DoF tibiofemoral joint, a 1 DoF patellofemoral joint, and a 1 DoF ankle joint. The motions of the lower limb were driven by 80 Hill-type muscle-tendon units of which the properties are based on the MRI data of young, physically active individuals (Rajagopal et al., 2016). The patellar tendon was modelled as an integral part of the quadriceps muscle-tendon unit. The force transmission mechanism between the quadriceps and patellar tendon was modelled through fixed path points at the superior and inferior border of the patella (Rajagopal et al., 2016). Motions of the trunk and arm segments were driven by ideal torque actuators. The spatial dimensions of the segments, the optimal muscle fibre lengths and tendon slack lengths in the generic model were scaled to the values of the individual participants using the marker data from a static calibration trial. The maximum isometric muscle forces were scaled to the participant's body mass and height (Handsfield et al., 2014).

Kinematic and GRF data were low-pass filtered using a 4th order Butterworth filter with a cut-off frequency of 6 Hz. The start and end of the squat movement were determined based on the velocity of the sacrum marker surpassing a threshold of 10% of the maximum velocity. Because of the joint constraints, the kinematics of the squats were determined using a global optimisation approach, minimising the weighted sum of squared differences between the measured marker positions and the marker positions of the model (Lu & O'Connor, 1999). The movement of the patella was not controlled by markers, but by a spline function with the tibiofemoral flexion angle as a determinative variable (Rajagopal et al., 2016). Since no marker data from the arms were collected, the upper limb segments of the model were fixed in a typical barbell back squat position: 45° of shoulder abduction and 90° of shoulder external rotation, elbow flexion and pronation. Next, net joint moments were calculated based on the kinematics results and the GRF data using the inverse dynamics tool. Forces of individual muscle-tendon units were calculated

with a static optimisation approach (Mokhtarzadeh et al., 2014), where the sum of the moments produced by the individual muscle-tendon units has to match the net joint moment and the sum of all muscle activations squared has to be minimal. The PTF was calculated as the scalar sum of forces produced by the four quadriceps muscle-tendon units. This approach was possible since the four muscle-tendon units of the quadriceps run in parallel at the level of the patellar tendon, and it has been previously used to calculate the PTF (Hagen, Van Rossom, et al., 2023; Hagen, Vanmechelen, et al., 2023). The peak PTF was normalised to the participant's body weight (BW).

4.2.5. Statistical analysis

For each participant, the average peak PTF value over the five trials per condition was taken as input for the statistical analysis. One participant was not able to perform any valid trial for the decline squat at 90% 1RM because of fatigue. Here, a data imputation technique was used by adding the average difference in peak PTF between the 80% 1RM and 90% 1RM decline squat for all other participants to the peak PTF of the 80% 1RM decline squat of this one participant. The statistical analysis was performed in the open-source software package SPM1D M0.4.9 (https://spm1d.org) (Pataky, 2012). The peak PTF data were normally distributed and the assumption of sphericity was not violated. Therefore, a parametric, zero-dimensional two-way repeated measures ANOVA's (mass * surface) was used. In case of a significant main effect for mass or an interaction effect, paired t-tests were used as post-hoc analyses to pinpoint the exact differences. The level of significance was set to 0.05.

4.3. Results

The five men and seven women included in this study were 24 ± 3 years old, had a body mass of 77 ± 18 kg and a body height of 175 ± 9 cm (mean ± standard deviation). The minimum and maximum 1RM values ranged between 30-110 kg on the decline board and 33-125 kg on level ground. Descriptive values for the masses of the external weights for each tested condition can be found in Table 4.1.

	Median (minimum-maximum) (kg)		
	70% 1RM	**80% 1RM**	**90% 1RM**
Decline board	27 (20-77)	31 (24-88)	35 (27-99)
Level ground	33 (23-87,5)	36 (26-100)	40 (30-112,5)

Table 4.1. *Median, minimum and maximum values for the mass of the external weights in each squat condition.*

Mean values and standard deviations for the different single-leg squat conditions and the results of the two-way repeated measures ANOVA for the peak PTF can be found in Table 4.2 and Figure 4.2. Only a significant main effect of the surface was found (F=6.731, p=0.025, partial eta squared=0.380) with higher peak PTF values on the decline board compared to level ground. The main effect for mass (F=2.566, p=0.100, partial eta squared=0.189) and the interaction effect (F=2.697, p=0.090, partial eta squared=0.197) were not statistically significant.

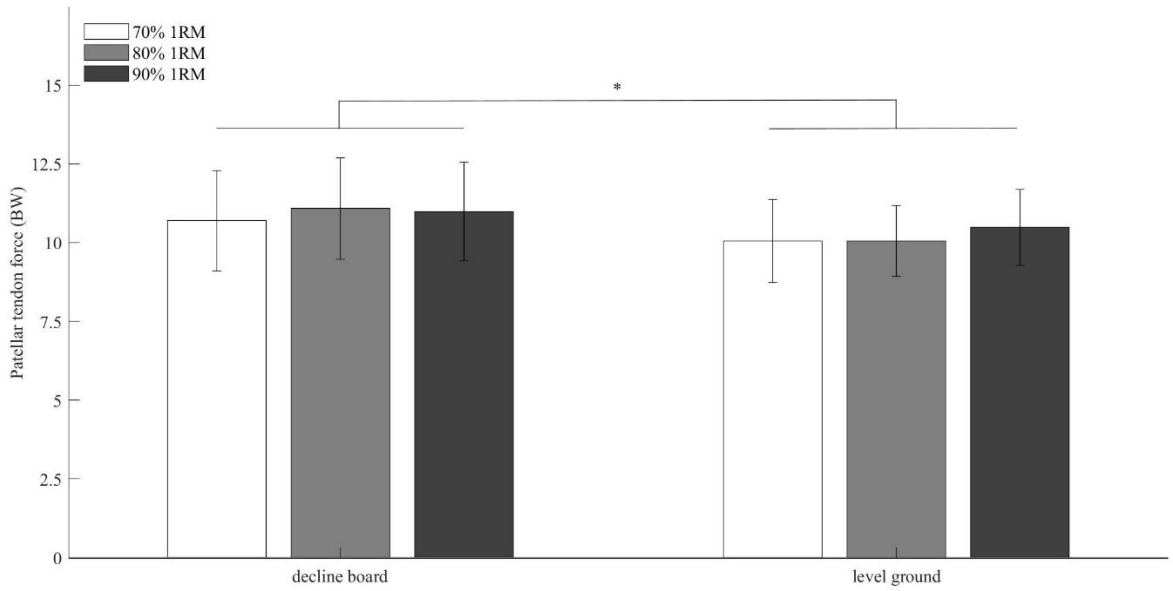

Figure 4.2. *Mean values ± standard deviations for peak patellar tendon force.*
** = p<0.05*

	Mean ± standard deviation (BW)		
	70% 1RM	**80% 1RM**	**90% 1RM**
Decline board	10.70 ± 1.59	11.08 ± 1.60	10.98 ± 1.56
Level ground	10.06 ± 1.32	10.05 ± 1.12	10.05 ± 1.20
	Two-way repeated measures ANOVA		
	Mass	**Surface**	**Mass*Surface**
F-value	2.566	6.731	2.697
p-value	0.100	0.025	0.090
Partial eta squared	0.189	0.380	0.197

Table 4.2. *Mean values ± standard deviations and results of the two-way repeated measures ANOVA for peak patellar tendon force.*

4.4. Discussion

The objective of this study was to investigate the influence of the mass of an external weight and the use of a decline board on the peak PTF during a heavy load single-leg squat. Both our original hypotheses were confirmed. There was no significant main effect for the mass of the external weights nor a significant interaction effect. In other words, both on the decline board and on level ground the peak PTF did not increase with increasing mass of the external weights from 70% to 90% of the 1RM. Furthermore, peak PTF values were higher on the decline board compared to level ground across all masses.

With no significant increase between 70% and 90% 1RM, the peak PTF followed a similar trend as other knee load parameters, such as the knee joint moment (Armstrong et al., 2022; Farris et al., 2016; Flanagan & Salem, 2008) or quadriceps muscle activity (Armstrong et al., 2022; McBride et al., 2010; van den Tillaar et al., 2019). Two factors may play a role in the origin of this steady state with increased mass. First, the masses are expressed as a percentage of the 1RM. A higher 1RM value also means a larger absolute separation between the mass conditions. In this group of participants, the median increase in the mass of the external weight per 10% 1RM was 3-4 kg, which can be insufficient to result in a tangible increase of the peak PTF. Second, the entire extensor kinetic chain is involved in the lifting of these increasingly heavier weights. Peak joint moments of the hip, knee and ankle can be found in the Supplementary analysis. Interestingly, the increase in joint moment was predominantly observed at the level of the hip extensors. Earlier research on the single-leg decline squat

suggested performing the squats with an upright trunk to put more stress on the knee extensors instead of the hip extensors (Jonsson & Alfredson, 2005; J. Zwerver et al., 2007). In this study, no instructions were given to the participants regarding the trunk position because it was not possible to maintain the upright trunk position during both the eccentric and concentric phase of a single-leg squat until 90° of knee flexion on both the decline and level surfaces. Furthermore, increased demands on the hip extensors instead of the knee extensors can be desirable since the importance of the kinetic chain and the inclusion of hip-specific exercises in PT exercises programs is increasingly recognised (Malliaras et al., 2015; Silva et al., 2015).

The peak PTF on the decline board was higher compared to level ground. The decline board has this effect on the peak PTF because of a shift in the GRF moment arm. Compared to level ground, the GRF moment arm increases at the level of the knee and decreases at the level of the hip and the ankle (Richards et al., 2016; J. Zwerver et al., 2007) (see also Supplementary analysis). Previous studies investigating the effect of a decline board typically performed their measurements with body mass alone or a relatively low external weight of 10 kg (Frohm et al., 2007; Kongsgaard et al., 2006; Richards et al., 2016; J. Zwerver et al., 2007). This study provides evidence that the decline board maintains its' effect on the peak PTF in heavy external load conditions as well. This is interesting since the 1RM on the decline board was lower than on level ground.

This study has some limitations that have to be discussed. First, it was not possible to standardise the contraction velocity because of the unstable nature of the single-leg squat, especially on the decline board. As a result, it would not be appropriate to calculate other relevant PTF parameters like the rate of PTF development or the PTF impulse (Kösters et al., 2014; Pearson et al., 2007; Tayfur et al., 2022). Differences in contraction velocity had no impact on the peak PTF however, since the single-leg squat with heavy loads is a slow movement and the peak PTF occurred around the turning point from eccentric to concentric movement. Furthermore, since the peak PTF occurred around the turning point from eccentric to concentric movement, the results of this study are also applicable to isometric or eccentrically accentuated variations of the single-leg squat. Second, it should be noted that the peak PTF was calculated instead of directly measured. Since direct measurement of the peak PTF would require the implantation of a buckle transducer or an optic fibre sensor (Roriz et al., 2014), advanced musculoskeletal modelling is currently the only method that can determine the peak PTF in a non-invasive way. However, the calculation of peak PTF values through musculoskeletal modelling comes with some assumptions and simplifications. Within the context of this study, the most important assumptions that need to be mentioned are the

patellofemoral kinematics that was driven by a spline function and the fixed patellar tendon length that was assumed by the static optimisation approach. While this could have generated an offset in the absolute peak PTF values, the statistical analysis was not affected since this study had a repeated measures design and the offset in the peak PTF was expected to be similar for all squat conditions within a participant. Third, this study included healthy participants who were familiar with heavy resistance training, while no patients with PT were included. This might limit the generalisability of this study to the prevention and not the rehabilitation of patellar tendinopathy. However, another study from our research group does provide evidence that peak PTF progressions over a large number of rehabilitation exercises are equivalent in patients with patellar tendinopathy and healthy individuals (study 4).

The results of this study are particularly interesting for patients with PT and healthy athletes who have no prior experience with heavy resistance training. This study has shown that peak PTF values are not significantly different between 70% and 90% of the 1RM during a single-leg squat and therefore that training at 70% of the 1RM should already be sufficient. It is therefore not necessary to let inexperienced athletes immediately train with external weights close to the 1RM, which would increase the risk of injury. Furthermore, patellar tendon load is higher on a decline board while the absolute 1RM weight in this condition is lower. Therefore, it can be concluded that it is possible to more safely expose athletes to increasing peak PTF loads during prevention or rehabilitation programs by performing the single-leg squats at 70% on level ground first, progressing onto a decline board with less weight.

4.5. Conclusion

This study provided evidence on PTF load progressions during variations of the heavy load single-leg squat. Despite a lower mass of the external weights, peak PTF was higher on a decline board compared to level ground. Increasing the mass of the external weights from 70% to 90% of the 1RM did not significantly increase the peak PTF.

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Supplementary analysis

The single leg squat is a weight lifting exercise that loads the entire lower limb extensor chain. Increases in the mass of the external weight are therefore distributed between the hip, knee and ankle extensors. However, the increased load is not necessarily distributed equally among these three joints. The objective of this supplementary analysis was to explore the evolution of the peak hip, knee and ankle extensor moment during the single leg squat on a decline board and on level ground with external weights increasing from 70% to 90% of the 1RM. Descriptive values are shown in Figure 4.3 below. Comparative statistics were not performed on these data.

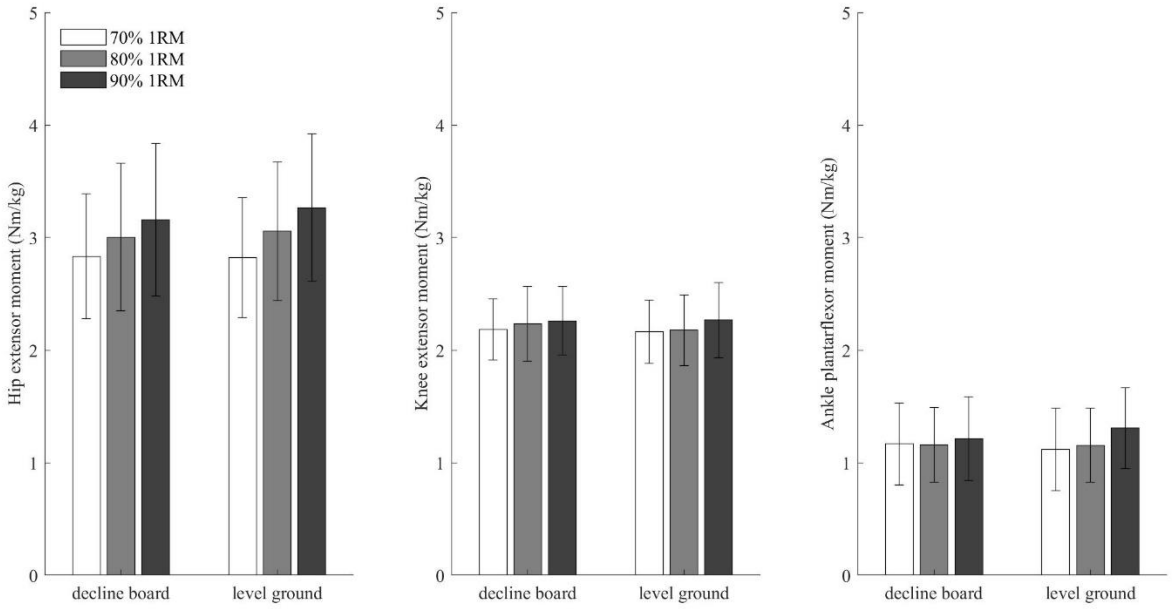

Figure 4.3. *Mean values ± standard deviations for the peak lower limb joint moments during the single leg squat on a decline board and level ground with external weights of 70%, 80% and 90% of the 1RM.*

Chapter 5

Study 4: Patellar tendon force progressions during rehabilitation exercises in patellar tendinopathy patients and healthy individuals: an equivalence study

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In preparation for submission

Abstract

Objectives

The objective of this study was to assess the degree of equivalence in the progression of peak patellar tendon forces (PTF) over a series of rehabilitation exercises in patients with patellar tendinopathy and healthy individuals.

Methods

Seven patients with patellar tendinopathy and seven healthy individuals participated in this study. Each participant performed 28 rehabilitation exercises while kinematics and ground reaction forces were measured. Peak PTF values were calculated through musculoskeletal modelling. Equivalence was tested by calculating the Pearson correlation coefficient for the mean values of all exercises.

Results

The patients with patellar tendinopathy executed the rehabilitation exercises with mild pain scores, while the healthy individuals performed all exercises pain-free. Nevertheless, the Pearson correlation coefficient for the mean values of all exercises was 0.980 ($p<0.001$), indicating near-perfect equivalence in the peak PTF progressions in both groups.

Conclusion

Peak PTF progressions are equivalent in patients with patellar tendinopathy and healthy individuals. Objectively quantified load progression principles for healthy individuals can be transferred to patients with patellar tendinopathy.

Key words

Patellar tendinopathy; exercise therapy; patellar tendon force; equivalence study

5.1 Introduction

Patellar tendinopathy (PT) is an overuse injury characterised by persistent patellar tendon pain and loss of function related to mechanical loading (Scott et al., 2020). This overuse injury is highly prevalent in the young, physically active population, especially in athletes involved in sprinting and jumping sports (Nutarelli et al., 2023). Interestingly, while physical activity with inadequate loading is seen as a causal factor for the development of PT, exercise therapy with optimal loading is seen as a primary treatment intervention (Cook & Purdam, 2009; Malliaras et al., 2015). In the rehabilitation of PT, mechanotherapy is a key principle. According to this principle, the load capacity of the injured tendon can be increased through the application of an optimal and progressively increasing dose of mechanical load (Khan & Scott, 2009). Hence, it is essential to provide physiotherapists with objective information regarding patellar tendon loading during physical activities and rehabilitation exercises.

Current scientific literature objectively quantifying patellar tendon forces (PTF) has frequently been conducted in healthy individuals instead of patients with PT (Elvin et al., 2009; Finni et al., 2000; Hagen, Vanmechelen, et al., 2023; Richards et al., 2016; Zellmer et al., 2019). Therefore, the question can be asked to what extent these results can be applied to patients with PT. For example, PTF progressions may be different in patients with PT due to pain-avoidance strategies in their movement pattern. However, studies comparing patients with PT and healthy individuals during jump-related tasks have rather focussed on finding differences between both groups as a primer for the development of injury prevention and return to sport tests (Bisseling et al., 2007; S. Edwards et al., 2010; Harris et al., 2020). Studies evaluating the equivalence of PTF progressions during exercise in patients with PT and healthy individuals remain a gap in the existing literature. Therefore, the objective of this study was to assess to what extent PTF progressions over a large number of rehabilitation exercises are equivalent in patients with PT and healthy individuals. The specific parameter of interest for this study, was the peak PTF. We hypothesised that the equivalence in peak PTF progressions between both groups would be large for the low load exercises, but that the peak PTF would increase at a slower rate in the patients with PT during the high load exercises.

5.2 Methods

5.2.1 Participants

Seven patients with PT and seven healthy individuals participated in this study. Patients with PT were recruited through the University Hospitals Leuven, Belgium and a network of local physiotherapists. Each patient had an official diagnosis of PT, made by a physician based on clinical examination. The diagnostic criterium for patellar tendinopathy is load-related pain localised at the inferior pole of the patella (Malliaras et al., 2015; Scott et al., 2020). Patellar tendon abnormality on medical imaging was not included as a diagnostic criterium since such abnormality can exist in absence of patellar tendon pain (Malliaras et al., 2015). The group included both acute and chronic patients with the time between the diagnosis and the data collection session altering between 1 month and 15 years. All patients with PT were actively involved in a rehabilitative exercise program at the moment of the data collection session. No patient had a history of infiltration or surgery at the level of the patellar tendon. The healthy individuals had no injury in the lower limb or trunk in the past six months. Both groups were matched for their sex and also body mass and body height were matched as good as possible. The demographic characteristics of each group can be found in Table 5.1. The study was approved by the ethical committee of UZ/KU Leuven, Belgium (reference number: S63980). All participants provided written informed consent before the start of data collection.

	PT patients (n=7)	**Healthy (n=7)**
Sex		
Male	4	4
Female	3	3
Age (years)	26 ± 6	23 ± 2
Body mass (kg)	69.1 ± 10.7	67.2 ± 10.3
Body height (m)	1.75 ± 0.10	1.74 ± 0.07
Score VISA-P	67 ± 19	97 ± 8

Table 5.1. *Mean values ± standard deviations for the demographic characteristics and scores on the VISA-P questionnaire of the patients with patellar tendinopathy (PT) and the healthy individuals.*

5.2.2 Data collection

Each participant was invited for one biomechanical data collection session during which he/she performed a series of rehabilitation exercises. The exercises consisted of variations of basic knee rehabilitation exercises (i.e. squats, lunges and backward step downs), sport-specific exercises (i.e. double-leg jumps and single-leg hops) and non-knee-specific exercises (i.e. walking, double-leg calf raises and single-leg romanian deadlifts), to cover a wide range of patellar tendon loads. An overview of the exercises can be found in Table 5.2 and the video of Supplementary analysis A. Participants performed five repetitions of each exercise.

Squat until 90° knee flexion
Double-leg squat body mass alone
Double-leg squat +10% extra body mass
Double-leg squat +20% extra body mass
Double-leg squat +30% extra body mass
Single-leg squat
Single-leg decline squat
Lunge 80% leg length
Forward lunge without step
Forward lunge with step
Lateral lunge without step
Backward step down
10 cm step height
20 cm step height
30 cm step height
Maximal effort double-leg jump
One vertical jump stiff-knee landing
One vertical jump squat landing
Two vertical jumps plyometric
One forward jump
Two forward jumps plyometric
Maximal effort single-leg hop
One vertical hop
Two vertical hops plyometric
One forward hop
Two forward hops plyometric
Non-knee-specific exercises
Walking
Double-leg calf raises
Single-leg romanian deadlift

Table 5.2. *List of exercises performed by the participants.*

The biomechanical data collection included three-dimensional kinematic data and ground reaction forces (GRF). Kinematic data were collected from the lower limbs and trunk (Malfait et al., 2014) at a sampling rate of 100Hz using 13 MX-T20 cameras of a passive marker-based system (Vicon, Oxford, UK). GRF data were collected at a sampling frequency of 1000Hz with two ground-embedded force plates (type OR6-7, AMTI, Watertown, MA, USA).

Pain scores were monitored throughout the data collection session. Before the start of the data collection, the participants filled in the VISA-P questionnaire (Zwerver et al., 2009) (Table 5.1). After each exercise, the participants scored the pain at the level of their patellar tendon on a 10-point numeric rating scale (Haefeli & Elfering, 2006). Both the exercise groups and the exercises within each group were randomised using an online randomization tool (www.randomizer.org) to prevent systematic order effects. In between exercises, participants were allowed to rest as long as they wanted. A five-minute warm-up on a stationary bike was provided before the start of the data collection session.

5.2.3 Data processing

The peak PTF was calculated through musculoskeletal modelling in OpenSim 3.3 (OpenSim, Stanford, CA, USA) (Delp et al., 2007; Seth et al., 2018) using the Catelli model (Catelli et al., 2019) which is – to our knowledge – the current best practice model for movements including deep knee and hip flexion angles. The lower limbs of this model consisted of a 6 DoF pelvis, a 3 DoF hip joint, a 1 DoF tibiofemoral joint, a 1 DoF patellofemoral joint, and a 1 DoF ankle joint. The motions of the lower limb were driven by 80 Hill-type muscle-tendon units of which the properties are based on the MRI data of young, physically active individuals (Rajagopal et al., 2016). The patellar tendon was modelled as an integral part of the quadriceps muscle-tendon unit. The force transmission mechanism between the quadriceps and patellar tendon was modelled through fixed path points at the superior and inferior border of the patella (Rajagopal et al., 2016). The data processing started by scaling the generic model to match the properties of each individual participant. The spatial dimensions of the bone segments, the optimal muscle fibre lengths and tendon slack lengths in the generic model were scaled based on the marker data from a static calibration trial. The maximum isometric forces of the muscle-tendon units in the model were scaled to the participant's body mass and height (Handsfield et al., 2014). For the jumping and hopping tasks, the maximum isometric forces were multiplied by a factor of two in order to avoid major reliance on reserve torque actuators during the high-speed concentric contractions at the level of the calf muscles (Kotsifaki et al., 2022).

Before entering the modelling workflow, kinematic and GRF data were low-pass filtered using a 4th order Butterworth filter with a cut-off frequency of 15 Hz for the jumping and hopping tasks and 6 Hz for the other rehabilitation exercises. The joint kinematics were determined using a global optimisation approach, minimising the weighted sum of squared differences between the measured marker positions and the marker positions of the model (Lu & O'Connor, 1999). The patellofemoral kinematics were not controlled by markers, but by a spline function with the tibiofemoral flexion angle as a determinative variable (Rajagopal et al., 2016). Next, the net joint moments were calculated based on the joint kinematics and the GRF data using inverse dynamics. Finally, individual muscle-tendon unit forces were calculated with a static optimisation approach (Mokhtarzadeh et al., 2014), where the sum of the moments produced by the individual muscle-tendon units has to match the net joint moment and the sum of all muscle activations squared has to be minimal. The PTF was calculated as the scalar sum of forces produced by the four quadriceps muscle-tendon units. This approach was possible since the four muscle-tendon units of the quadriceps run in parallel at the level of the patellar tendon, and has been used in previous research to calculate the PTF (Hagen, Van Rossom, et al., 2023; Hagen, Vanmechelen, et al., 2023).

For the patients with unilateral PT, the data of the symptomatic leg was included and in patients with bilateral PT, the data of the most painful leg was included. In the group of matched healthy individuals, the leg was chosen according to limb dominance. For those patients who had their dominant leg injured, the dominant leg was selected in their matched healthy individuals as well and vice versa for the non-dominant leg. The peak PTF value over the entire trial was identified and normalised to the participant's body weight (BW). The average peak PTF value of the five trials per participant per exercise was used for the statistical analysis.

5.2.4 Statistical analysis

Descriptive values in the form of mean values and standard deviations of the peak PTF per exercise were calculated for each group. In the jump and landing tasks without plyometrics, the descriptive values were calculated separately for the push-off and landing phases. For some exercises, the data of one or more participants had to be removed due to inconsistencies. Two data points were removed from the group of patients with PT while 12 datapoints were removed from the group of healthy individuals.

Next, the Pearson correlation coefficient was calculated for the mean values of all exercises to determine the degree of linear association between the peak PTF values of both groups. For the rehabilitation of PT, it is important to know whether or not the peak PTF progressed at a similar pace in both groups. Absolute agreement in the peak PTF between both groups is less relevant. Therefore, the correlation coefficient was chosen over the intraclass correlation coefficient (Portney & Watkins, 2014b). The minimal value for a significant correlation was set to 0.05.

5.3 Results

Mean values and standard deviations of the peak PTF and pain scores per exercise for each group can be found in Table 5.3 and Figure 5.1. The correlation coefficient for the mean values of all exercises was 0.980 ($p<0.001$) (Figure 5.2). Some exercises fall slightly below the equivalence line, meaning that the mean PTF values of these exercises were slightly higher in the healthy individuals compared to the patients with PT. The mean values for the non-knee-specific exercises (double-leg calf raises, single-leg romanian deadlifts and walking) were relatively wide apart from the other rehabilitation exercises, potentially inflating the correlation coefficient artificially. Hence, the correlation coefficient was recalculated without the non-knee-specific exercises. Yet, even with these exercises removed, the correlation coefficient remained very high: 0.933 ($p<0.001$).

Exercise	**PT patients**	**Healthy**	**Overall**
Squat until 90° knee flexion			
Double-leg squat body mass alone	6.72 ± 2.03	7.47 ± 2.58	7.09 ± 2.26
	(0 ± 1)	(0 ± 0)	
Double-leg squat +10% extra body mass	8.66 ± 2.52	9.11 ± 2.24	8.89 ± 2.30
	(1 ± 1)	(0 ± 0)	
Double-leg squat +20% extra body mass	8.93 ± 2.26	9.09 ± 2.39	9.01 ± 2.24
	(1 ± 1)	(0 ± 0)	
Double-leg squat +30% extra body mass	10.07 ± 2.73	10.11 ± 2.42	10.09 ± 2.48
	(1 ± 1)	(0 ± 0)	
Single-leg squat	10.18 ± 1.13	10.24 ± 0.42	10.21 ± 0.82
	(1 ± 1)	(0 ± 0)	
Single-leg decline squat	10.80 ± 1.69	10.82 ± 0.39	10.81 ± 1.22
	(2 ± 2)	(0 ± 0)	

Lunge 80% leg length			
Forward lunge without step	8.19 ± 1.67	7.62 ± 1.03	7.90 ± 1.36
	(1 ± 1)	(0 ± 0)	
Forward lunge with step	8.76 ± 0.65	8.57 ± 1.15	8.67 ± 0.88
	(1 ± 1)	(0 ± 0)	
Lateral lunge without step	8.21 ± 2.27	8.82 ± 2.50	8.54 ± 2.32
	(0 ± 0)	(0 ± 0)	
Backward step down			
10 cm step height	4.85 ± 1.16	4.39 ± 1.08	4.62 ± 1.11
	(0 ± 1)	(0 ± 0)	
20 cm step height	7.03 ± 1.38	6.57 ± 1.66	6.80 ± 1.49
	(1 ± 1)	(0 ± 0)	
30 cm step height	8.66 ± 1.68	8.22 ± 1.33	8.44 ± 1.47
	(1 ± 1)	(0 ± 0)	
Maximal effort double-leg jump			
One vertical jump push-off	8.68 ± 1.21	10.37 ± 2.32	9.52 ± 1.98
	(1 ± 1)	(0 ± 0)	
One vertical jump stiff-knee landing	5.76 ± 1.83	7.04 ± 2.14	6.40 ± 2.03
One vertical jump squat landing	(0 ± 1)	(0 ± 0)	
	8.88 ± 1.11	9.57 ± 1.39	9.22 ± 1.26
Two vertical jumps plyometric	(1 ± 1)	(0 ± 0)	
	10.39 ± 1.32	11.11 ± 1.76	10.72 ± 1.52
One forward jump push-off	(1 ± 1)	(0 ± 0)	
	8.82 ± 1.84	9.63 ± 3.50	9.16 ± 2.54
One forward jump landing	(1 ± 1)	(0 ± 0)	
	10.24 ± 1.98	11.60 ± 3.40	10.73 ± 2.51
Two forward jumps plyometric	(1 ± 1)	(0 ± 0)	
	12.46 ± 1.18	12.51 ± 2.31	12.48 ± 1.56
	(1 ± 1)	(0 ± 0)	

Maximal effort single-leg hop			
One vertical hop push-off	9.29 ± 1.95 (2 ± 2)	9.16 ± 1.48 (0 ± 0)	9.23 ± 1.67
One vertical hop landing	7.90 ± 1.68 (2 ± 2)	8.36 ± 1.30 (0 ± 0)	8.13 ± 1.47
Two vertical hops plyometric	8.67 ± 1.05 (1 ± 2)	9.47 ± 1.99 (0 ± 0)	9.07 ± 1.58
One forward hop push-off	8.08 ± 1.02 (2 ± 3)	9.31 ± 1.61 (0 ± 0)	8.65 ± 1.42
One forward hop landing	9.21 ± 1.23 (2 ± 3)	9.49 ± 1.96 (0 ± 0)	9.34 ± 1.54
Two forward hops plyometric	9.79 ± 1.18 (2 ± 2)	10.27 ± 1.72 (0 ± 0)	9.99 ± 1.37
Non-knee-specific exercises			
Walking	2.02 ± 0.43 (0 ± 0)	2.32 ± 0.57 (0 ± 0)	2.18 ± 0.51
Double-leg calf raises	0.34 ± 0.20 (0 ± 0)	0.29 ± 0.14 (0 ± 0)	0.32 ± 0.17
Single-leg romanian deadlift	1.54 ± 0.32 (0 ± 0)	1.09 ± 0.69 (0 ± 0)	1.29 ± 0.58

Table 5.3. *Mean values ± standard deviations for the peak patellar tendon force in times body weight of each exercise for the patients with patellar tendinopathy (PT) and the healthy individuals. Mean values ± standard deviations for the pain scores at the level of the patellar tendon on a 10-point numeric rating scale are mentioned between brackets.*

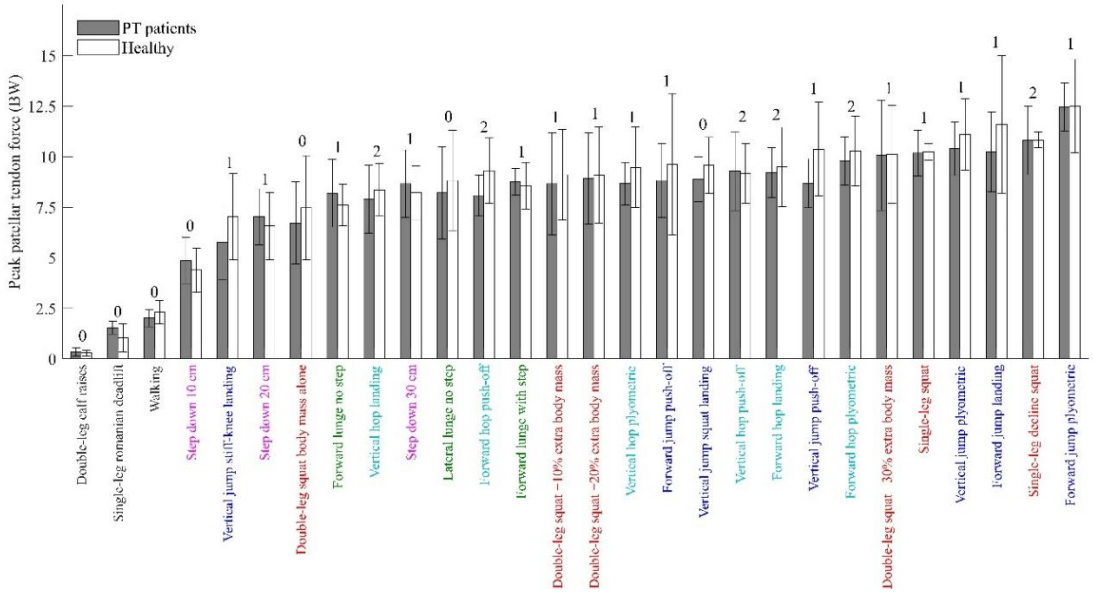

Figure 5.1. *Bar plot with the mean peak patellar tendon forces in times body weight (BW) of each exercise for the patients with patellar tendinopathy (PT) and the healthy individuals. The exercises were ranked from left to right based on their overall mean peak patellar tendon forces from both groups (see Table 5.3). Mean pain scores at the level of the patellar tendon on a 10-point numeric rating scale in the patients with PT are mentioned above each bar. Mean pain scores of the healthy individuals were always 0.*

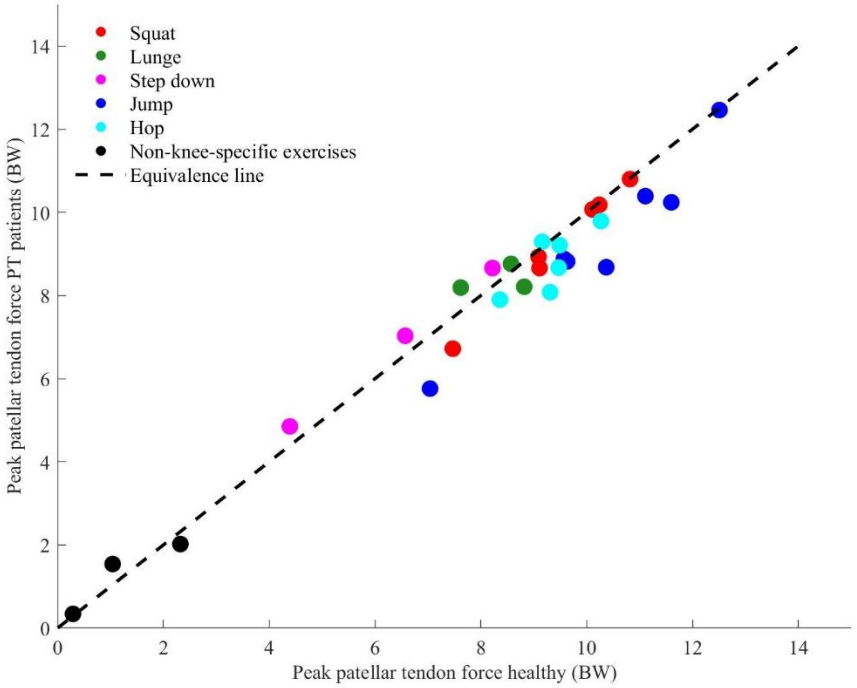

Figure 5.2. *Scatter plot of the mean peak patellar tendon forces in times body weight (BW) for the patients with patellar tendinopathy (PT) and the healthy individuals. The dashed equivalence line indicates equal absolute peak patellar tendon force values in both groups.*

5.4 Discussion

The objective of this study was to assess to what extent peak PTF progresses equally in patients with PT and in healthy individuals across a wide range of rehabilitation exercises. Peak PTF values were objectively quantified over a total of 28 exercises, covering the entire spectrum from low to high load exercises. The results of this study showed a very high correlation between the peak PTF progression of the patient and healthy groups during these exercises. Earlier studies investigating PTF progressions during squats (Kongsgaard et al., 2006; Richards et al., 2016; J. Zwerver et al., 2007), lunges (Hagen, Van Rossom, et al., 2023; Zellmer et al., 2019) or jumps (Elvin et al., 2009; Finni et al., 2000) have only been conducted with healthy individuals. The results of the current study provide supportive evidence that these load progression principles for healthy individuals can be transferred to patients with PT.

A hypothetical reason why the peak PTF progression could have been different in patients with PT, was a load reduction strategy in order to avoid pain. If the peak PTF in patients with PT would increase at a similar rate as in healthy individuals during the low load exercises, but at a slower rate in the high load exercises, peak PTF progressions would be different in both groups. In this study, the patients executed the high load rehabilitation exercises with mild pain scores, as typically prescribed in rehabilitation programs for PT (Escriche-Escuder et al., 2020). Despite these mild pain scores, peak PTF values for the squats, lunges and step downs remained close to the equivalence line. Some jumping tasks were farther away from the equivalence line with lower values in the patients with PT, which was probably caused by slightly lower jump heights and distances (Elvin et al., 2009; I. Janssen et al., 2015) (see Supplementary analysis B). However, whether the reduced jump performance in the patients with PT was a consequence of the pathology and pain, cannot be concluded from this cross-sectional study. Furthermore, the pain scores during the jumping and hopping tasks were not distinctly higher compared to the other exercises. Finally, the statistical analysis identified a very high linear correlation, indicating that the overall peak PTF progression was not affected.

The descriptive values reported in this study could potentially provide a starting point for the creation of a normative dataset for peak PTF progressions over a large variation of rehabilitation exercises. Creating such a dataset is necessary because measurement tools that objectively quantify PTF in a physiotherapy practice setting have not been developed yet (Verheul et al., 2020). With this dataset, physiotherapists would have an important tool to optimise their progressive tendon loading programs. However, it should be mentioned that the peak PTF values in this study were calculated with musculoskeletal modelling, which has its limitations.

A particular limitation that has to be discussed in the context of this study, is the patellofemoral kinematics since it influences the patellar tendon moment arm. In Supplementary analysis C – which compared the calculated muscle activations from the musculoskeletal modelling approach against measured muscle activations from electromyography (EMG) – a clear indication was given that the patellar tendon moment arm of the musculoskeletal model was likely too short in the more flexed knee joint angles. Since the amount of underestimation in the patellar tendon moment arm was angle-specific, ranking of the rehabilitation exercises from light to heavy was affected, with exercises involving more flexed knee joint angles shifting towards a lower peak PTF rank. Another important limitation of this approach is the fact that static optimisation does not include the contribution of passive elastic elements. In order to compensate for this, the maximum isometric muscle forces of the model had to be multiplied by two during the jump and hop tasks. Because of these limitations, validation of the calculated peak PTF values is warranted before a normative dataset based on this musculoskeletal modelling approach is adopted. Ideally, the calculated peak PTF values are compared against directly measured peak PTF values from a buckle transducer or optic fibre sensor. Alternatively, an indirect comparison can be made between the calculated muscle activations from the musculoskeletal modelling approach and the measured muscle activations from EMG.

This study has two limitations that need to be addressed. First, it should be mentioned that the peak PTF is not the only relevant load parameter in the context of PT. Other load parameters like the rate of PTF development (Kösters et al., 2014) and PTF impulse (Firminger et al., 2020; Pearson et al., 2007) have also shown their relevance. During the strength training phase of a progressive tendon loading program, the peak PTF is progressively increased while the PTF impulse is kept high and the rate of PTF development relatively low. During the next phase, jumping tasks are progressively re-introduced, increasing the rate of PTF development and decreasing the PTF impulse (Malliaras et al., 2015). Investigating the rate of PTF development and PTF impulse in this study would not be appropriate, since it would require a different data collection protocol with standardisation of and variations in the time under tension and task execution velocity. Namely, the time under tension and task execution velocity of a rehabilitation exercise can change throughout the rehabilitation program depending on the specific objective. For example, on the one hand a squat can be performed during the strength training phase with a fixed time under tension of two seconds eccentric muscle contractions, one second isometric muscle contractions and two seconds concentric muscle contractions. One the other hand a squat at maximal contraction velocity can be used as a transitional exercise during the progressive re-introduction of jumping tasks. Future research can therefore extend

upon the current study by providing additional data on peak PTF values, the rate of PTF development and PTF impulse. Second, the number of participants included was small: seven patients with PT and seven healthy individuals. This is because there is a trade-off between the number of rehabilitation exercises and the number of participants that can be measured, given the extensive amount of time the advanced biomechanical data collection, processing and analysis take. As mentioned previously, the primary objective of this study was to assess to what extent peak PTF progresses equally in patients with PT and in healthy individuals across a wide range of rehabilitation exercises. Since the correlation coefficient required a sufficient amount of rehabilitation exercises, the number of participants that could be measured and analysed, was limited. A sample size of seven participants per group to calculate the mean values for a regression analysis over 28 different rehabilitation exercises, is small. Nevertheless, this group of participants can be considered representative since the standard deviations for most rehabilitation exercises are relatively small and similar in both groups. Furthermore, this study included patients with a wide variety of symptom duration and severity, making the results of this study more generalisable.

5.5 Conclusion

This study provides evidence that peak PTF during rehabilitation exercises progress in an equivalent way in patients with PT and healthy individuals. Therefore, earlier discovered peak PTF progression principles for healthy individuals can safely be transferred to patients with PT. Furthermore, this study could provide a starting point for the creation of a normative dataset for peak PTF progressions over a large variation of rehabilitation exercises.

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Supplementary analysis A

Video of the rehabilitation exercises:

https://drive.google.com/file/d/1Nlby_v7CAonI_IlXDDA_cGLAi--DbmAa/view?usp=sharing

Supplementary analysis B

Exercise	PT patients	Healthy
Maximal effort double-leg jump		
One vertical jump push-off	0.38 ± 0.07	0.39 ± 0.09
One vertical jump stiff-knee landing	0.38 ± 0.07	0.38 ± 0.09
One vertical jump squat landing	0.38 ± 0.07	0.39 ± 0.09
Two vertical jumps plyometric	0.37 ± 0.07	0.36 ± 0.08
One forward jump push-off	1.66 ± 0.36	1.93 ± 0.42
One forward jump landing	1.71 ± 0.41	1.76 ± 0.45
Two forward jumps plyometric	1.69 ± 0.43	1.75 ± 0.46
Maximal effort single-leg hop		
One vertical hop push-off	0.29 ± 0.05	0.26 ± 0.04
One vertical hop landing	0.29 ± 0.05	0.26 ± 0.04
Two vertical hops plyometric	0.25 ± 0.08	0.24 ± 0.05
One forward hop push-off	1.33 ± 0.28	1.71 ± 0.39
One forward hop landing	1.31 ± 0.31	1.55 ± 0.42
Two forward hops plyometric	1.32 ± 0.31	1.47 ± 0.37

Table 5.4. *Mean values ± standard deviations of the jump heights and distances (in meter) for the patients with patellar tendinopathy (PT) and the healthy individuals.*

Supplementary analysis C

The peak patellar tendon forces (PTF) in this equivalence study were calculated using musculoskeletal modelling with a static optimisation approach, but this comes with some assumptions and simplifications. Therefore, validation of the peak PTF calculated by this musculoskeletal modelling approach is warranted. Unfortunately, no study has directly validated different approaches for PTF calculations against golden standard measures such as buckle transducers or optic fibre sensors (Roriz et al., 2014). Only two studies have reported in vivo PTF measurements during jumping tasks with optic fibre sensors (Elvin et al., 2009; Finni et al., 2000). First, Finni et al. (2000) et al. measured peak PTF during the push-off phase of squat jumps and countermovement jumps at moderate intensity. For the squat jumps with a jump height of 0.16 ± 0.04m (mean ± standard deviation), the peak PTF was 3200 ± 1463N. During the countermovement jumps with a jump height of 0.18 ± 0.05m, the peak PTF was 4297 ± 2147N. Later, Elvin et al. (2009) measured peak PTF during the push-off and landing phases of countermovement jumps with increasing intensity. Jump heights in this study ranged from 0.35m to 0.58m with a mean height of 0.43m. During the push-off, the mean peak PTF was 4590N and the maximum peak PTF was 7500N. During the landing, the mean peak PTF was 4000N and the maximum peak PTF was 5500N.

Although none of these data can be directly compared against the results of this equivalence study, a basic comparison can be made when the results of the vertical jump variations are reconverted to peak PTF values in Newton. Furthermore, the peak PTF was also calculated according to a second, frequently used approach: dividing the knee extensor moment by the angle-specific moment arm according to the equations of Herzog & Read (1993). Table 5.5 shows that on the one hand the equation moment divided by moment arm is an underestimation of the peak PTF compared to the data from optic fibre sensors as described above. This is no surprise since the co-contractions of the hamstrings and gastrocnemius muscles are not taken into account in this equation. The results of the static optimisation approach on the other hand seem an overestimation compared to the data from optic fibre sensors.

	Moment / moment arm	Static optimisation
One vertical jump push-off	2106 ± 564	6367 ± 1510
One vertical jump stiff-knee landing	2518 ± 554	4159 ± 949
One vertical jump squat landing	2221 ± 791	6137 ± 1073
Two vertical jumps plyometric	2671 ± 839	7245 ± 1270

Table 5.5. *Mean values ± standard deviations for the peak patellar tendon force in Newton according to two different calculation approaches*.

A second verification step that can be performed for the static optimisation approach, is a comparison between the calculated muscle activations from the static optimisation and measured muscle activations from electromyography (EMG). This comparison was performed for five patients with PT and three healthy individuals during all rehabilitation exercises (Figure 5.3). The comparison shows that the calculated muscle activations of the vastus lateralis and medialis are higher than what was experimentally observed through EMG. Interestingly, the discrepancy was more pronounced for exercises with higher knee flexion angles. For example, during the backward step down, the discrepancy between the static optimisation activations and the EMG activations increased as the peak knee flexion increased (Figure 5.4). The comparison between the static optimisation activations and the EMG activations over the entire movement pattern for the other exercises can be found at:
https://drive.google.com/drive/folders/1aHN2q3DCsqzG3E6hEnF6g8tchRC5eDBW?usp=drive_link

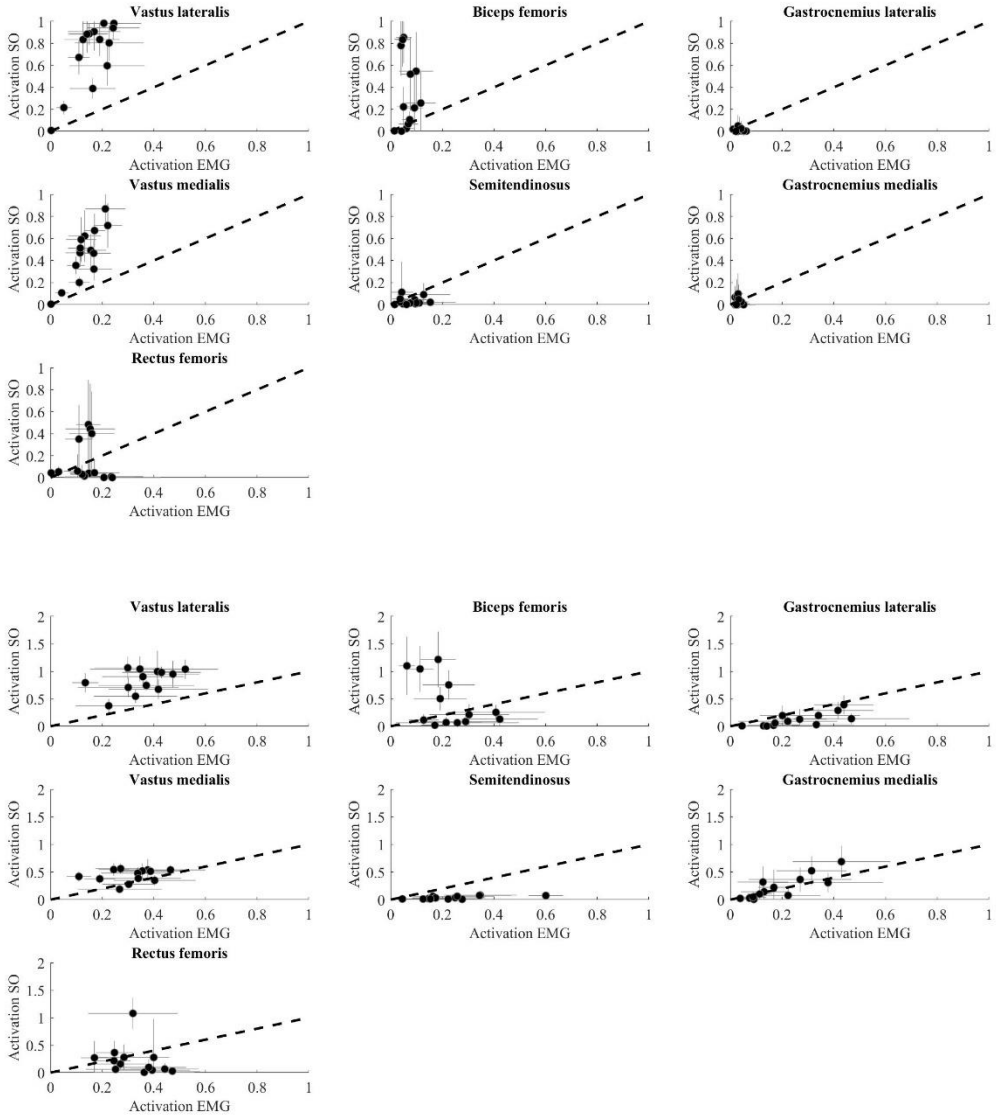

Figure 5.3. *Comparison between the calculated muscle activations from the static optimisation (SO) and measured muscle activations from electromyography (EMG) at the moment of peak patellar tendon force for five patients with PT and three healthy individuals during all rehabilitation exercises. Dots indicate the mean values and error bars indicate the standard deviations. The dashed equivalence line indicates no difference between the SO and EMG activations. Top: squats, lunges, backward step downs and non-knee-specific exercises. Bottom: double-leg jumps and single-leg hops. Since the maximum isometric muscle forces of the model were doubled for the analysis of the double-leg jumps and single-leg hops, the static optimisation activations were doubled for these exercises as well.*

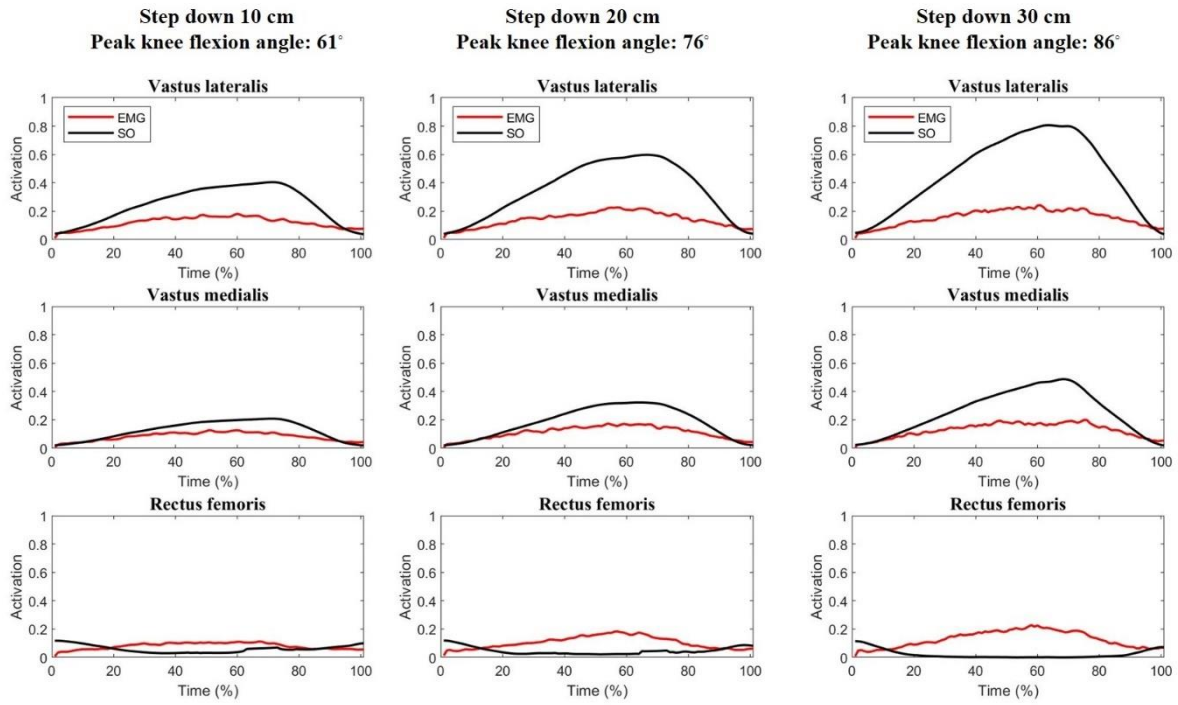

Figure 5.4. *Comparison between the mean calculated muscle activations from the static optimisation (SO) in black and the mean measured muscle activations from electromyography (EMG) in red during the backward step down at 10, 20 and 30 cm. Mean values for the peak knee flexion angle are shown for each step height.*

A potential reason for the discrepancy between the muscle activations from static optimisation and EMG can be found in the patellar tendon moment arm. The patellar tendon moment arm of the musculoskeletal model was verified against the cadaveric study of Buford et al. (1997). Interestingly, these values for the patellar tendon moment arm are distinctly lower in more flexed knee angles compared to the patellar tendon moment arm values reported by Herzog & Read (1993) (Figure 5.5). Additionally, a systematic review by Tsaopoulos et al. (2006) demonstrated that the patellar tendon moment arm values reported by Buford et al. (1997) are an outlier and that the values reported by Herzog & Read (1993) follow the general trend. In order to verify whether the patellar tendon moment arm of the model was the cause of the discrepancy between the static optimisation and EMG activations, the angle-specific ratio between the patellar tendon moment arm of the model and Herzog & Read (1993) was calculated. Next, the muscle activation patterns from static optimisation were multiplied by this angle-specific ratio. When these corrected static optimisation muscle activations were again compared against the muscle activations from EMG, the data showed better correspondence (Figure 5.6 and Figure 5.7). The comparison between the corrected static optimisation activations and the EMG activations over the entire movement pattern for each rehabilitation exercise can be found at:
https://drive.google.com/drive/folders/1u7AAwQbn06zbZ_7pbyv2WqQYx-ZWnHrD?usp=drive_link

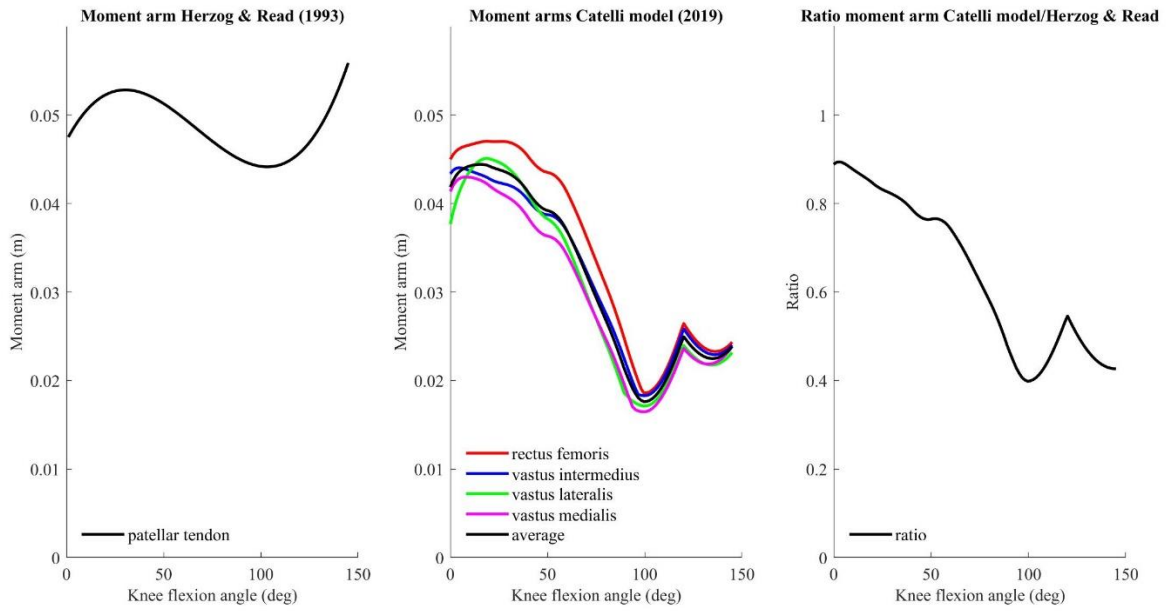

Figure 5.5. *Differences in the patellar tendon moment arm between the study of Herzog & Reid (1993) and the Catelli model (2019). The moment arms from the Catelli model were based on the study of Buford et al. (1997).*

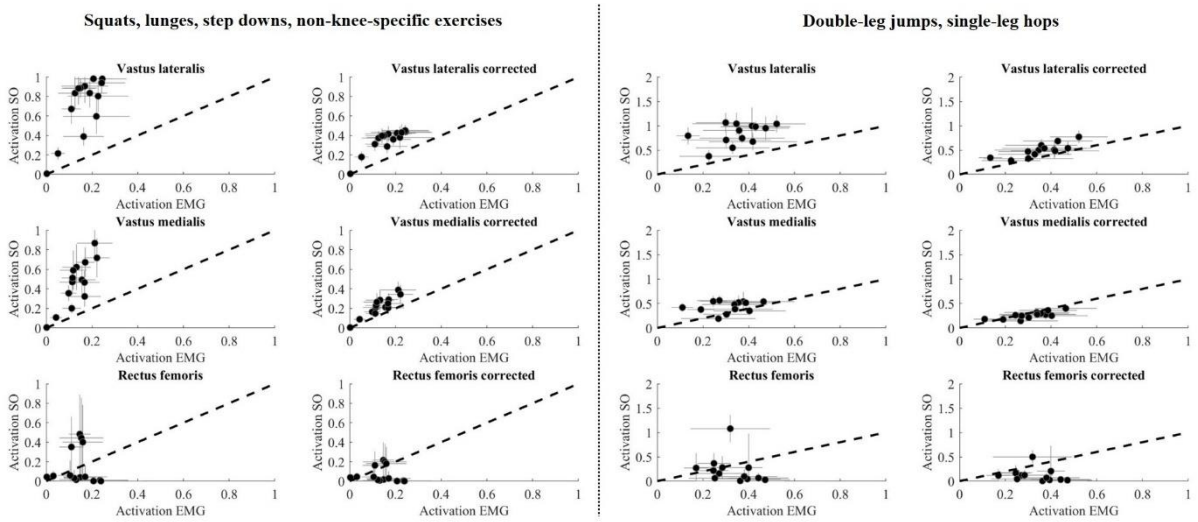

Figure 5.6. *Comparison between the calculated quadriceps activations from the static optimisation (SO) and the measured quadriceps activations from electromyography (EMG) at the moment of peak patellar tendon force before and after correction for the difference in patellar tendon moment arms. Dots indicate the mean values and error bars indicate the standard deviations. The dashed equivalence line indicates no difference between the SO and EMG activations. Left: squats, lunges, backward step downs and non-knee-specific exercises. Right: double-leg jumps and single-leg hops. Since the maximum isometric muscle forces of the model were doubled for the analysis of the double-leg jumps and single-leg hops, the static optimisation activations were doubled for these exercises as well.*

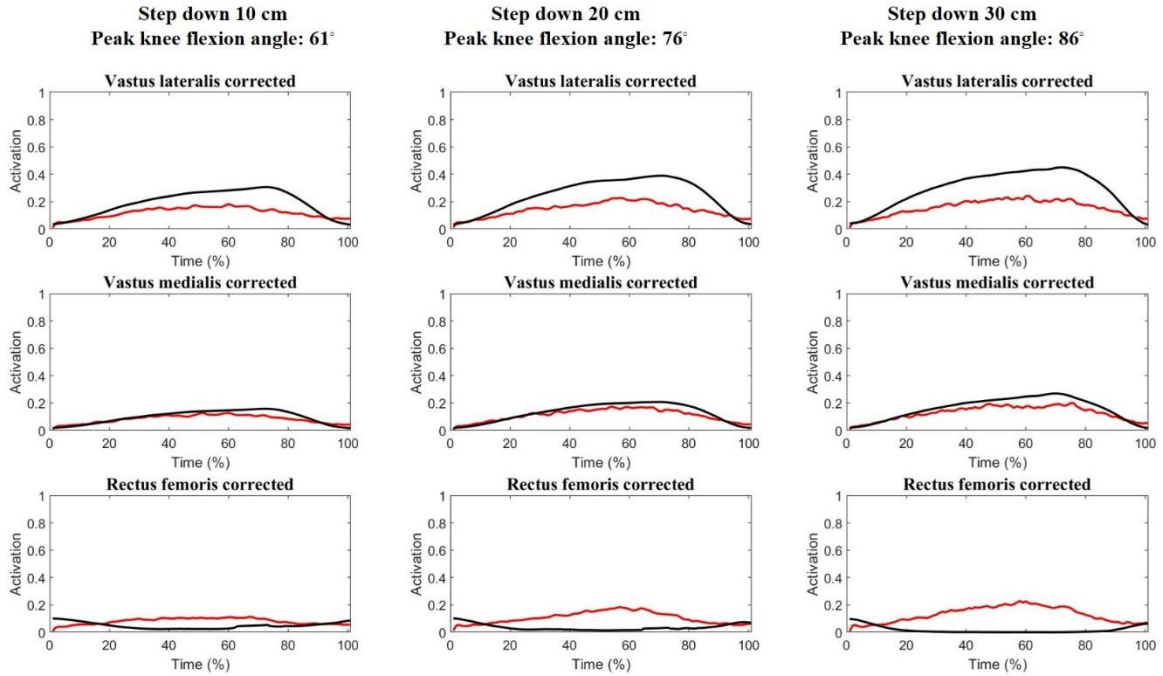

Figure 5.7. *Comparison between the mean calculated muscle activations from the static optimisation (SO) in black and the mean measured muscle activations from electromyography (EMG) in red during the backward step down at 10, 20 and 30 cm after correction for the difference in patellar tendon moment arms. Mean values for the peak knee flexion angle are shown for each step height.*

To demonstrate the effect of the differences in the patellar tendon moment arm on the results of this study, the peak PTF was recalculated with the angle-specific ratio applied. The correlation between corrected mean peak PTF values of the patients with PT and healthy individuals remained at 0.980 (<0.001). Therefore, it can be concluded that the equivalence in peak PTF progressions between both groups is not affected by the differences in the patellar tendon moment arm. However, the differences in the patellar tendon moment arm did affect the descriptive peak PTF values (Table 5.6). Not only were the mean values and standard deviations lower for the corrected peak PTF, but also the ranking of the rehabilitation exercises from light to heavy was affected. Based on the comparison with EMG, the corrected ranking seems a better estimation of the true ranking compared to the original ranking. However, this correction is only an approximation. The correct procedure would be to adjust the patellar tendon moment arm of the musculoskeletal model – probably by adjusting the spline function for the patellofemoral kinematics – and run the static optimisation again. Furthermore, the comparison between the corrected static optimisation activations and the EMG was good for these eight participants, but potential patellofemoral malalignment or maltracking and their effects on the patellar tendon moment arm might influence these results in another group of participants

(Kujala et al., 1987). Direct measurement of the patellofemoral kinematics and patellar tendon moment arm through dynamic medical imaging was not available for this study. Hence, the patellofemoral kinematics of the model was controlled by a generic spline function (Rajagopal et al., 2016; van Eijden et al., 1986).

In conclusion, there is still uncertainty involved with objectively quantifying the peak PTF progressions over a broad variety of rehabilitation exercises with different knee flexion angles. Future research could measure the angle-specific patellar tendon moment arm of their participants with dynamic medical imaging and adjust their musculoskeletal models accordingly before performing the static optimisation analysis. Furthermore, direct comparison between the PTF from musculoskeletal modelling and from direct measurements (e.g. using optic fibre sensors) is warranted.

	Mean ± SD		**Ranking**	
Exercise	**Original**	**Corrected**	**Original**	**Corrected**
Squat until 90° knee flexion				
Double-leg squat body mass alone	7.09 ± 2.26	3.39 ± 0.93	7	5
Double-leg squat +10% extra body mass	8.89 ± 2.30	4.10 ± 0.96	14	8.5
Double-leg squat +20% extra body mass	9.01 ± 2.24	4.17 ± 0.91	15	11
Double-leg squat +30% extra body mass	10.09 ± 2.48	4.65 ± 1.02	23	14
Single-leg squat	10.21 ± 0.82	4.95 ± 0.27	24	17
Single-leg decline squat	10.81 ± 1.22	5.12 ± 0.31	27	18
Lunge 80% leg length				
Forward lunge without step	7.90 ± 1.36	3.51 ± 0.57	8	6
Forward lunge with step	8.67 ± 0.88	3.88 ± 0.46	13	7
Lateral lunge without step	8.54 ± 2.32	4.10 ± 0.71	11	8.5
Backward step down				
10 cm step height	4.62 ± 1.11	3.32 ± 0.66	4	4
20 cm step height	6.80 ± 1.49	4.13 ± 0.62	6	10
30 cm step height	8.44 ± 1.47	4.50 ± 0.50	10	12

Maximal effort double-leg jump				
One vertical jump push-off	9.52 ± 1.98	4.85 ± 0.81	21	16
One vertical jump stiff-knee landing	6.40 ± 2.03	4.80 ± 1.29	5	15
One vertical jump squat landing	9.22 ± 1.26	4.55 ± 0.74	18	13
Two vertical jumps plyometric	10.72 ± 1.52	5.29 ± 0.79	25	20
One forward jump push-off	9.16 ± 2.54	5.14 ± 1.32	17	19
One forward jump landing	10.73 ± 2.51	5.69 ± 1.05	26	21
Two forward jumps plyometric	12.48 ± 1.56	6.37 ± 0.84	28	26
Maximal effort single-leg hop				
One vertical hop push-off	9.23 ± 1.67	6.00 ± 0.58	19	23
One vertical hop landing	8.13 ± 1.47	5.89 ± 0.85	9	22
Two vertical hops plyometric	9.07 ± 1.58	6.08 ± 0.88	16	24
One forward hop push-off	8.65 ± 1.42	6.32 ± 1.16	12	25
One forward hop landing	9.34 ± 1.54	6.67 ± 1.23	20	27
Two forward hops plyometric	9.99 ± 1.37	6.97 ± 0.70	22	28
Non-knee-specific exercises				
Walking	2.18 ± 0.51	1.80 ± 0.41	3	3
Double-leg calf raises	0.32 ± 0.17	0.28 ± 0.14	1	1
Single-leg romanian deadlift	1.29 ± 0.58	1.06 ± 0.45	2	2

Table 5.6. *Mean values ± standard deviations for the original and corrected peak patellar tendon force in times body weight of all participants. The exercises were ranked from light (1) to heavy (28) based on the mean peak patellar tendon force.*

Chapter 6

Study 5: Can physiotherapists accurately estimate patellar tendon forces during rehabilitation exercises? A biomechanics-informed survey

Michiel Hagen, Pieter-Jan Peeters, Stein Van Noten, Danilo S. Catelli, Dieter Van Assche, Sabine Verschueren, Jos Vanrenterghem

In preparation for submission

Abstract

Objectives

Exercise therapy with progressive patellar tendon loading is key for the rehabilitation of patellar tendinopathy. Unfortunately, objective quantification of patellar tendon forces in a physiotherapy practice is not yet possible. Physiotherapists rely on visual inspection and clinical reasoning to estimate these forces. The objective of this study was to examine whether physiotherapists can accurately estimate patellar tendon forces during rehabilitation exercises.

Methods

The patellar tendon force of 19 rehabilitation exercises was objectively quantified using musculoskeletal modelling. Videos of these exercises were implemented in an online survey. Participants were asked to score each exercise for peak patellar tendon force on a 100-point numerical rating scale. Fifty actively practicing physiotherapists and 50 final year master students filled out the survey. The overall accuracy was calculated using the intraclass correlation coefficient $(ICC)_{2,k}$ both for the entire group and for subgroups based on years of experience and specialisation. The accuracy of individual exercises was determined using t-tests.

Results

The $ICC_{2,k}$ for the entire group was 0.895 (0.727 – 0.960) which is good to excellent. Across all subgroups, both based on years of experience and specialisation, $ICC_{2,k}$ values were good to excellent as well. Nonetheless, the peak PTF of six exercises was overestimated and the peak PTF of four exercises was underestimated.

Conclusion

Physiotherapists have a good to excellent overall ability to estimate peak patellar tendon forces compared to the current best practice musculoskeletal modelling approach. This estimating ability was similar across subgroups based on years of experience and specialisation.

Key words

Patellar tendinopathy; exercise therapy; reliability; visual rating

6.1 Introduction

Patellar tendinopathy is a common and disabling injury among the young, physically active population (Nutarelli et al., 2023). This overuse injury is defined as persistent patellar tendon pain and loss of function related to mechanical loading (Scott et al., 2020). The rehabilitation of patellar tendinopathy is based on the principle of mechanotherapy: stimulating the adaptive capacity of an injured structure through an optimal and progressively increasing dose of mechanical load (Cook & Purdam, 2009; Khan & Scott, 2009). In consequence, load management and exercise therapy with gradual progression of patellar tendon load are considered key elements in the rehabilitation of patellar tendinopathy (Breda et al., 2021; Malliaras et al., 2015).

Unfortunately, the objective quantification of patellar tendon forces (PTF) requires expensive, advanced and time-consuming data collection and analysis techniques, such as 3D motion capture, force plates and musculoskeletal modelling. It is therefore not possible to objectively quantify PTF in a physiotherapy practice setting to create progressive exercise programs for individual patients (Van Rossom et al., 2018; Verheul et al., 2020). At present, physiotherapists typically rely on their visual inspection and clinical reasoning skills to estimate PTF.

Visual inspection has already proven to be a reliable tool for measures such as range of motion (Russo et al., 2018) or movement quality (Lindblom et al., 2021). Furthermore, the influence of experience on the reliability of visual inspection for these measurements has also been investigated (Gulgin & Hoogenboom, 2014; Weeks et al., 2012; Whatman et al., 2012). However, neither the reliability of visual inspection nor the influence of experience has been investigated for the ability to estimate musculoskeletal loading, let alone PTF specifically. Therefore, the objective of this study was to examine whether physiotherapists can accurately estimate PTF during rehabilitation exercises based on visual inspection. A secondary objective was to investigate the influence of experience on their accuracy.

6.2 Methods

First, reference data on PTF during rehabilitation exercises were collected. For this, one healthy individual (male, 24 years of age, body mass 81.2kg, body height 1.78m) performed twice a series of 19 rehabilitation exercises. These rehabilitation exercises were chosen to cover a wide range of PTF and included: the push-off and landing of vertical and forward double-leg jumps and single-leg hops, double-leg and single-leg squats on level ground and a decline board, forward and lateral lunges with and without step, backward step-down, walking, single-leg romanian deadlift and double-leg calf raises (see Video in Supplementary analysis). The first time, the rehabilitation exercises were performed in a biomechanics laboratory to objectively quantify PTF. During the second session, the same rehabilitation exercises were carried out in a clinical setting and videos of the frontal and sagittal plane were recorded. These videos were implemented in an online survey and sent to actively practicing physiotherapists and final year master students in physiotherapy (see details below).

6.2.1 Objective quantification

During the data collection session in the biomechanics lab, kinematic and ground reaction force (GRF) data were collected while the participant performed the rehabilitation exercises. Kinematic data were collected for the lower limb and trunk (Malfait et al., 2014) at a sampling rate of 100Hz using a passive marker-based motion capture system (type MX-T20, Vicon, Oxford, UK). Simultaneously, GRF data were collected at a sampling rate of 1000Hz from two ground-embedded force plates (type OR6-7, AMTI, Watertown, MA, USA). Kinematic and GRF data were low-pass filtered using a 4th-order Butterworth filter with a cut-off frequency of 15 Hz for the jumping/hopping tasks and 6 Hz for the other rehabilitation exercises.

The specific load parameter of interest in this study was the peak PTF. This load parameter was calculated through musculoskeletal modelling in OpenSim 3.3 (OpenSim, Stanford, CA, USA) (Delp et al., 2007). The Catelli model (Catelli et al., 2019), created based on MRI data of young and active individuals (Rajagopal et al., 2016), was used as generic model. First, this generic model was scaled to match the anthropometric values of the participant using marker data from a static calibration trial. The maximum isometric forces of the muscle-tendon units in the model were scaled to the participant's body mass and height (Handsfield et al., 2014). For the jumping/hopping tasks, the maximum isometric forces were multiplied by two to avoid major reliance on reserve torque actuators during the high-speed concentric contractions (Kotsifaki et al., 2022). Next, the joint kinematics were determined using a global optimisation approach,

minimising the weighted sum of squared differences between the measured marker positions and the marker positions of the model (Lu & O'Connor, 1999). Net joint moments were then calculated based on the kinematics results and the GRF data using inverse dynamics. Finally, individual muscle-tendon unit forces were calculated with a static optimisation approach (Mokhtarzadeh et al., 2014), where the sum of the moments produced by the individual muscle-tendon units has to match the net joint moment, and the sum of all muscle activations squared has to be minimal. The PTF was calculated as the scalar sum of forces produced by the four quadriceps muscle-tendon units (Hagen, Van Rossom, et al., 2023; Hagen, Vanmechelen, et al., 2023).

6.2.2 Survey

An online survey was created on the Qualtrics platform (Qualtrics International Inc.; Provo, Utah, USA). In this survey, the physiotherapists were shown anonymised video clips of the frontal and sagittal plane in which the healthy participant performed the rehabilitation exercises. This time, the videos were recorded in a clinical setting and the participant was wearing regular sport clothing (see Video in Supplementary analysis).

The physiotherapists rated each exercise for peak PTF on a 0-100 numeric rating scale. A score of 0 was defined as 'total absence of patellar tendon force (0 times body weight)' whereas a score of 100 was defined as the 'maximum patellar tendon force across all exercises (10.9 times body weight)'. The objective peak PTF of one exercise – i.e. sit-to-stand on a chair with a height of 45cm – was given as a reference load to the participants. The peak PTF during this activity was 60% (6.5 times body weight).

6.2.3 Physiotherapists

A total of 100 participants from Flanders, Belgium completed the survey. The group consisted of 50 graduated, actively practicing, musculoskeletal physiotherapists and 50 final year master students in physiotherapy majoring in musculoskeletal rehabilitation across different Flemish universities. In order to investigate the influence of experience on a physiotherapist's load estimating accuracy, the total group was divided into subgroups in two different ways: once in three groups based on years of experience (students – five years or less actively practicing physiotherapists – six years or more actively practicing physiotherapists) and once in two groups based on specialisation (sports physical therapy – manual therapy or no specialisation). An overview of the characteristics of each subgroup can be found in Table 6.1. Before the start of the survey, each participant digitally signed a written informed consent. The study was approved by the ethical committee of UZ/KU Leuven (reference number S63980).

	Sports physical therapy	**Manual therapy or musculoskeletal general**	**Total**
Students	31	19	50
≤ 5 years actively practicing physiotherapist	13	11	24
> 5 years actively practicing physiotherapists	15	11	26
Total	59	41	

Table 6.1. *Number of physiotherapists per subgroup.*

6.2.4 Statistical analysis

The results of the survey were analysed in SPSS v28 (IBM, New York, USA). The intraclass correlation coefficient (ICC) $_{2,k}$ was used to estimate the overall accuracy of the subjective peak PTF estimations (Portney & Watkins, 2014b). Next to the $ICC_{2,k}$ for the entire group, separate $ICC_{2,k}$ values were calculated for each subgroup. A post-hoc analysis was performed to quantify the accuracy at the level of the individual exercises. One-sample t-tests were performed on the difference scores between the subjective and objective peak PTF values to investigate which exercises specifically were estimated correctly, overestimated or underestimated. A subjective estimation was considered incorrect if two criteria were met. First, the difference score was significantly different from zero. To avoid type 1 errors because of the large number of exercises, the critical alpha value was set to 0.01. Second, Cohen's d effect size measure was larger than 0.5, which is arbitrarily considered as medium. This criterium was chosen to avoid clinically irrelevant differences from being marked as incorrect. The post-hoc analysis was performed on the difference scores of the entire group. A post-hoc analysis for a specific subgroup was only performed if the 95% confidence interval of the $ICC_{2,k}$ for this subgroup did not overlap with the $ICC_{2,k}$ of the entire group.

6.3 Results

$ICC_{2,k}$ values for the entire group and each subgroup can be found in Table 6.2. The $ICC_{2,k}$ for the entire group was 0.895 (0.727 – 0.960) which is arbitrarily considered as good to excellent. Furthermore, there were no differences between the subgroups, neither based on years of experience nor based on specialisation. The 95% confidence intervals overlapped between all subgroups.

Group	$ICC_{2,k}$ (95% CI)
Entire group	0.895 (0.727-0.960)
Students	0.886 (0.701-0.956)
≤ 5 years actively practicing physiotherapist	0.889 (0.709-0.957)
> 5 years actively practicing physiotherapists	0.909 (0.761-0.965)
Sports physical therapy	0.898 (0.732-0.961)
Manual therapy or musculoskeletal general	0.892 (0.721-0.958)

Table 6.2. *Intraclass correlation coefficients for the entire group of physiotherapists and for each subgroup based on years of experience and specialisation.*
ICC = intraclass correlation coefficient

Post-hoc analysis of the entire group for the individual exercises revealed that the peak PTF of nine exercises were estimated correctly, six exercises were overestimated and four exercises were underestimated (see Table 6.3 and Figure 6.1). The overestimated exercises included calf raises ($p<0.001$, $d=1.114$), single-leg romanian deadlift ($p<0.001$, $d=1.408$), walking ($p<0.001$, $d=0.52$), double-leg squat ($p<0.001$, $d=0.836$), stiff-knee landing of double-leg vertical jump ($p<0.001$, $d=1.629$) and landing of single-leg vertical hop ($p<0.001$, $d=0.543$). The peak PTF of the lateral lunge ($p<0.001$, $d=-1.339$), landing of the double-leg forward jump ($p<0.001$ $d=-0.566$), single-leg squat ($p<0.001$, $d=-1,620$) and the single-leg decline squat ($p<0.001$, $d=-1,192$) was underestimated.

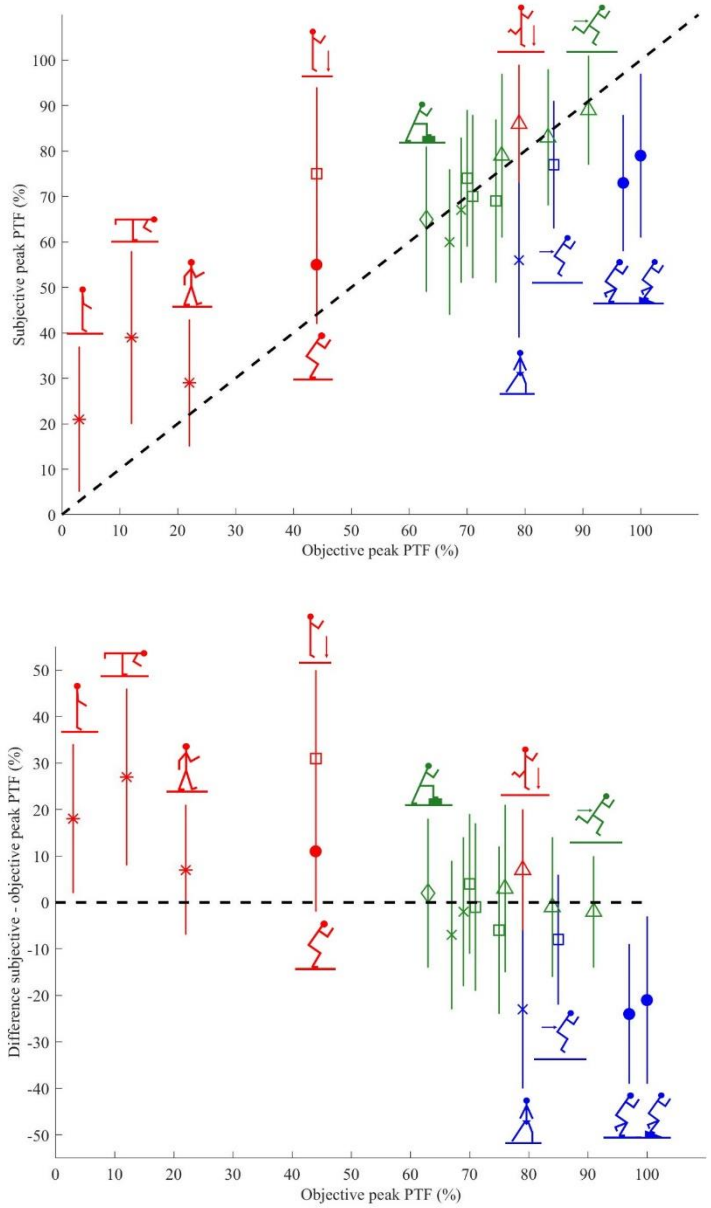

Figure 6.1. *Top: scatter plot of the objective PTF values and mean values ± standard deviations of the subjectively estimated peak PTF of the entire group for each exercise. Bottom: Mean values ± standard deviations for the difference scores between the subjective estimations of the entire group and the objective peak PTF values for each exercise.*

The dashed equivalence line indicates no difference between the subjective and objective peak PTF values. Exercises that were estimated correctly are indicated in green. Exercises indicated in red were overestimated and exercises indicated in blue were underestimated.

** = non-knee specific exercises*

• = squats

x = lunges

□ = jumps

Δ = hops

Exercise	Objective peak PTF % max	Mean subjective peak PTF % max	Difference subjective – objective peak PTF Mean ± SD
Single-leg decline squat	100	79	-21 ± 18
Single-leg squat	97	73	-24 ± 15
Forward hop landing	91	89	-2 ± 12
Forward jump landing	85	77	-8 ± 14
Vertical hop push-off	84	83	-1 ± 15
Vertical hop landing	79	86	7 ± 13
Lateral lunge without step	79	56	-23 ± 17
Forward hop push-off	76	79	3 ± 18
Vertical jump push-off	75	69	-6 ± 18
Forward jump push-off	71	70	-1 ± 18

Vertical jump squat landing	70	74	4 ± 15
Forward lunge with step	69	67	-2 ± 16
Forward lunge without step	67	60	-7 ± 16
Backward step down (20cm)	63	65	2 ± 16
Vertical jump stiff-knee landing	44	75	31 ± 19
Double-leg squat	44	55	11 ± 13
Walking	22	29	7 ± 14
Single-leg romanian deadlift	12	39	27 ± 19
Double-leg calf raise	3	21	18 ± 16

Table 6.3. *Objective peak patellar tendon force, overall mean subjectively estimated peak patellar tendon forces and difference scores (mean ± standard deviations) between the subjective estimations of the entire group and the objective peak patellar tendon force values for each exercise individually.*

Exercises that were estimated correctly are indicated in green. Exercises indicated in red were overestimated and exercises indicated in blue were underestimated. Exercises were considered over- or underestimated when $p<0.01$ and Cohen's $d>0.5$.

PTF = patellar tendon force

6.4 Discussion

The objective of this study was to examine whether physiotherapists can accurately estimate PTF during rehabilitation exercises based on visual inspection and to investigate the influence of experience on their accuracy. The results of this study show that the overall PTF-estimating accuracy of musculoskeletal physiotherapists is good to excellent compared to advanced musculoskeletal modelling. Furthermore, the accuracy of all subgroups based on years of experience or specialisation was similar. In other words, final year master students in both sports physical therapy and manual therapy have a good to excellent overall PTF-estimating accuracy by the time they graduate and this accuracy is maintained throughout their career as actively practicing physiotherapists. This is important since gradual progression of patellar tendon load is a key element in the rehabilitation of patellar tendinopathy (Breda et al., 2021; Malliaras et al., 2015) and measurement tools to objectively quantify patellar tendon loads in clinical practice are not yet available.

Although the $ICC_{2,k}$ indicated a good to excellent overall accuracy, the standard deviations at the level of the individual exercises can be considered large. Furthermore, the post-hoc analysis still identified systematic over- and underestimations for certain individual exercises. In general, exercises with low peak PTF were overestimated and exercises with a high peak PTF were underestimated. A few exercises deserve additional attention, starting with the single-leg squats on level ground and the decline board. The latter is typically prescribed in eccentric exercise programs (Jonsson & Alfredson, 2005) and as a pain-provocation exercise to check whether a patient with patellar tendinopathy tolerated the load of the exercise program (Malliaras et al., 2015). The physiotherapists recognised the high peak PTF of this exercise, by scoring their peak PTF similar to jumping and hopping tasks. However, most physiotherapists did not identify the single-leg decline squat as the heaviest exercise with a score of 100%. With overall mean percentages of 79% for the single-leg decline squat and 73% for the single-leg squat on level ground, these exercises have been ranked as the fifth and ninth exercise in a ranking from high to low. A second important group of exercises that need to be discussed, are the jumping and hopping tasks since patellar tendinopathy is highly prevalent in sports involving jumping (Nutarelli et al., 2023). The physiotherapists estimated the peak PTF of most jumping and hopping tasks correctly. Additionally, the landing of the forward jump and vertical hop were under- and overestimated, respectively, by less than 10% on average. However, the peak PTF of the vertical jump with stiff-knee landing was overestimated by 31% on average. Here, the physiotherapists did not recognise the fact that the stiff-knee landing generated only

a small external moment arm towards knee flexion, and therefore a relatively low peak PTF. Two final exercises that need to be discussed, are the double-leg calf raise and single-leg romanian deadlift. The involvement of the lower limb kinetic chain in the rehabilitation of patellar tendinopathy is increasingly recognised (Malliaras et al., 2015; Silva et al., 2015), so it is important that the peak PTF during ankle and hip specific exercises can be estimated correctly. The physiotherapists correctly identified the double-leg calf raise and single-leg romanian deadlift as the exercises with the lowest peak PTF, however the percentage peak PTF was largely overestimated. This could mean that physiotherapists are too cautious to implement these exercises in early-stage exercise programs. However, the objective peak PTF values show that there is no reason to be cautious with these exercises.

Some limitations of the current study and consequent recommendations for future research have to be discussed. First, all physiotherapists were shown videos of only one healthy individual performing the rehabilitation exercises. Hence, the question can be asked whether physiotherapists would estimate the peak PTF differently if they saw another person with a different movement pattern perform the same rehabilitation exercises, especially a patient with patellar tendinopathy. It should be mentioned though that the objective peak PTF values of the individual in this study are similar to that of a group of young individuals both with and without patellar tendinopathy (study 4). Second, only Flemish physiotherapists filled out the survey. Physiotherapy is well developed in this region with high-quality educational programs at the master level including biomechanical courses. It should be possible to generalise the results of this study to other regions and countries with a well-developed physiotherapy system, but the generalisability of this study to regions and countries where the physiotherapy system is still in development may be a topic for future research (Fidelis et al., 2023; Thornton et al., 2023). Third, the peak PTF estimations of the physiotherapists were verified against objectively quantified peak PTF values from musculoskeletal modelling. The musculoskeletal modelling approach of this study is considered the current best practice approach with regard to the specific characteristics of the individual and the exercises. Nevertheless, this approach has its own limitations in terms of accuracy as well. Supplementary analysis C of study 4 identified a discrepancy between the calculated muscle activations from the musculoskeletal modelling approach and measured muscle activations from electromyography, probably caused by an inaccuracy of the length of the patellar tendon moment arm. Since the magnitude of this inaccuracy was angle-specific, the ranking of the rehabilitation exercises from light to heavy was affected. Therefore, the PTF estimating accuracy can be compared against the current best practice modelling approach from this study, but future studies should verify these results once

more refined approaches for the objective quantification of PTF become available. In this context, it should also be mentioned that the minimal clinically important difference for the peak PTF is not yet known (Portney & Watkins, 2014b). Consequently, it is not known how accurate any peak PTF estimation or calculation should be in order to be considered a truly reliable tool in a physiotherapy practice. Hence, the criteria for the overall and exercise-specific peak PTF estimating accuracy were set based on conventionally accepted arbitrary values. Finally, it should be recognised that the current study is a simplification of the reality. Creating a rehabilitation program for patients with patellar tendinopathy, requires to take other load parameters into account as well. This includes structure-specific load parameters, such as the rate of PTF development or the PTF impulse (Firminger et al., 2020; Kösters et al., 2014; Pearson et al., 2007), but also external load parameters such as the total volume per session or the acute-chronic workload ratio (Gabbett, 2016). The future development of clinical measurement tools could assist the physiotherapist to put this complex puzzle together. Additionally, future measurement tools could have the advantage that they can be used by the patients themselves outside the clinical practice, allowing a more ecologically valid analysis of patellar tendon loading (Verheul et al., 2020).

6.5 Conclusion

Physiotherapists demonstrate a good to excellent ability to estimate peak PTF based on visual inspection, keeping in mind that this was in comparison to the current best practice musculoskeletal modelling approach and over a large variety of rehabilitation exercises. There was no difference in accuracy across subgroups based on years of clinical experience and specialisations. However, there are still important overestimations and underestimations at the level of individual exercises, which show that there is still room for improvement in the PTF estimating accuracy of physiotherapists. Nevertheless, this study suggests that patients with patellar tendinopathy can be confident in the PTF estimating capacity of their physiotherapist in the current absence of objective measurement tools in the physiotherapy practice.

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Supplementary analysis

Video of the rehabilitation exercises used in the survey:
https://drive.google.com/file/d/1fUq8UvE4sIFAKc06iTiMmis6ohiYBv49/view?usp=sharing

Chapter 7

Study 6: Are peak patellar tendon force estimations from kinematics and kinetics viable in the clinic? An exploratory analysis

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In preparation for submission

Abstract

Objectives

There is a lack of measurement tools that can objectively quantify peak patellar tendon forces in clinical practice. An important step towards the development of such tools, is the identification of proxy signals. The objective of this study was to explore the potential of peak patellar tendon force estimations based on sagittal plane kinematic and kinetic parameters for multiple rehabilitation exercises.

Methods

Kinematics and kinetic data were collected from nine patients with patellar tendinopathy and seven healthy individuals for 28 rehabilitation exercises. Peak patellar tendon forces were calculated through musculoskeletal modelling. The proxy signal identification occurred in three steps. The exercises were first separated into categories based on visual inspection of scatter plots. Next, two forms of stepwise linear regression were performed per category: basic regression with each signal as independent variable and principal component regression. Each regression form was performed on the kinematic data alone and on the kinematic and kinetic data combined.

Results

Four relevant exercise categories were identified (squats/lunges/step downs – jumping/hopping push-off – jumping/hopping landing – jumping/hopping plyometric). The adjusted R^2 values of basic regression analyses from kinematic data alone, varied between 0.398 and 0.663 ($p<0.001$). Adding the kinetic data improved all basic regression equations, except for the plyometric exercises. Principal component regression did not improve the adjusted R^2 values.

Conclusion

There is a potential for peak patellar tendon force estimations from sagittal plane kinematics and kinetics across several common rehabilitation exercises. Researchers and engineers can use these findings to develop future clinical measurement tools.

Key words

Patellar tendon force; proxy signal; regression; principal component analysis

7.1 Introduction

Patellar tendinopathy is a common overuse injury, especially in young people practicing sports that involve sprinting and jumping elements. Prevalence rates across different sports in non-elite and elite athletes are reported to be 19.9% and 16.7% respectively (Nutarelli et al., 2023). A key element in the rehabilitation of patellar tendinopathy is exercise therapy with an optimal and progressively increasing load on the patellar tendon to increase the load capacity of this structure (Cook & Purdam, 2009; Khan & Scott, 2009; Malliaras et al., 2015). To create such a progressive tendon loading program, objective information regarding patellar tendon load progressions during rehabilitation exercises is key. Unfortunately, it is not yet possible to objectively quantify patellar tendon forces (PTF) in a clinical practice setting (Van Rossom et al., 2018; Verheul et al., 2020). The reason for this lack of clinical measurement tools is twofold. First, the objective quantification of PTF currently requires expensive equipment, such as three-dimensional opto-electronic motion capture systems and force plates. Second, data processing typically demands advanced and time-consuming methods, including the use of specialised software and advanced musculoskeletal modelling. Consequently, physiotherapists have to rely on scientific studies reporting PTF values during rehabilitation exercises and their visual inspection skills to estimate PTF for individual patients. The development of clinical measurement tools for the objective quantification of PTF could therefore mean an important advancement in the rehabilitation of patellar tendinopathy.

An important step towards the development of clinical measurement tools, is the identification of proxy signals. These proxy signals are signals that vary in accordance with the actual load parameter of interest and can be measured with non-invasive, low-cost technologies. For example, it would be a large step forward if clinicians could reasonably track PTF progressions based on sagittal plane kinematic parameters from two-dimensional motion capture, since this would only require one video-camera. Earlier studies have tried to identify such proxy signals for PTF parameters during running (Brund et al., 2021) and jump landings (Ina Janssen et al., 2013) specifically. However, the identification of discrete groups of proxy signals for PTF parameters over multiple rehabilitation exercises combined, has not yet been attempted. The objective of the current study was therefore to take initial steps towards the identification of potential proxy signals for PTF parameters that can be used for multiple rehabilitation exercises. More specifically, the potential of peak PTF estimations based on sagittal plane kinematics of the lower limb and trunk was explored and compared against estimations with lower limb joint kinetics as additional input signals.

7.2 Methods

7.2.1 Data collection and analysis for peak PTF

The current exploratory study is a secondary analysis of a recently collected dataset of objectively quantified peak PTF values (study 4). In the original study, seven patients with patellar tendinopathy and seven healthy individuals were asked to perform a series of 28 rehabilitation exercises. Fourteen trials were discarded because of technical issues. For this proxy signal identification study, two more patients with patellar tendinopathy were included, resulting in a total of nine patients with patellar tendinopathy (26 ± 6 years old, 71.4 ± 10.3kg, 1.78 ± 0.11m, VISA-P score of 68 ± 17) and seven healthy individuals (23 ± 2 years old, 67.2 ± 10.3kg, 1.74 ± 0.07m, VISA-P score of 97 ± 8). One of the newly added patients did not perform any hop task and the other patient did not perform the plyometric variation of the vertical hop. This resulted in a total of 440 valid trials. The study was approved by the ethical committee of UZ/KU Leuven, Belgium (reference number: S63980) and all participants provided written informed consent before the start of data collection.

Data collection and analysis methods have been described in detail previously (study 4). In brief, participants were invited for one biomechanical data collection session during which they performed a series of rehabilitation exercises (see Table 7.1 and Supplementary analysis A). During the data collection session, three-dimensional kinematic data of the lower limb and trunk (Malfait et al., 2014) and ground reaction force data were collected. Both kinematic and force data were low-pass filtered using a 4th order Butterworth filter with a cut-off frequency of 15 Hz for the jumping/hopping tasks and 6 Hz for the other rehabilitation exercises. Joint angles, joint moment and peak PTF values for each exercise were then calculated in OpenSim 3.3 (OpenSim, Stanford, CA, USA) (Delp et al., 2007) using the Catelli model (Catelli et al., 2019). The peak PTF of each exercise was calculated through a static optimisation approach (Mokhtarzadeh et al., 2014).

Squat until 90° knee flexion
Double-leg squat body mass alone
Double-leg squat +10% extra body mass
Double-leg squat +20% extra body mass
Double-leg squat +30% extra body mass
Single-leg squat
Single-leg decline squat
Lunge 80% leg length
Forward lunge without step
Forward lunge with step
Lateral lunge without step
Backward step down
10 cm step height
20 cm step height
30 cm step height
Maximal effort double-leg jump
One vertical jump stiff-knee landing
One vertical jump squat landing
Two vertical jumps plyometric
One forward jump
Two forward jumps plyometric
Maximal effort single-leg hop
One vertical hop
Two vertical hops plyometric
One forward hop
Two forward hops plyometric
Non-knee-specific exercises
Walking
Double-leg calf raises
Single-leg romanian deadlift

Table 7.1. *List of exercises performed by the participants.*

7.2.2 Proxy signal identification

The signals included in this exploratory analysis, were the sagittal plane kinematics of the ankle, knee, hip and trunk – including joint angles, angular velocities and angular accelerations – and sagittal plane joint kinetics of the lower limb. For the patients with unilateral PT, the data of the symptomatic leg was included while for patients with bilateral PT, the data of the most painful leg was included. For the healthy individuals, the leg was chosen based on limb dominance: in those patients who had their dominant injured, the dominant leg was selected in their matched

healthy individuals as well and vice versa for the non-dominant leg. Given the large variety of exercises and the fact that peak PTF is zero-dimensional (i.e. a discrete value), key discrete data points were extracted from the sagittal plane kinematic and kinetic profile. More specifically, peak joint angles, velocities and accelerations for flexion and extension of the ankle, knee, hip and trunk and peak internal extensor moments for the ankle, knee and hip were determined. Furthermore, the joint angular velocity and acceleration curves were full-wave rectified and average flexion/extension velocities and accelerations over the entire curves were calculated.

The exploratory analysis itself consisted of three steps (see Figure 7.1). The first step was to identify relevant categories of rehabilitation exercises within the dataset. This step was undertaken to prevent that a clearly distinguishable relationship between the peak PTF and a proxy signal for one category of exercises would be occluded by a non-relationship between the peak PTF and that same proxy signal for another category of exercises. The identification of relevant exercise categories was performed based on a visual inspection of the scatter plots depicting the associations between the peak PTF and the candidate proxy signals.

In the second step, proxy signals were identified per category of exercises using a stepwise linear regression analysis (Portney & Watkins, 2014a). Because of the large number of input variables, the regression analysis was performed in two different ways. The first method was to perform a basic stepwise linear regression per category with each candidate proxy signal as a separate independent variable. This basic stepwise linear regression analysis was performed in Matlab R2020b (The Mathworks, Natick, MA, USA) using a built-in function. The criterium for a candidate proxy signal to enter the regression equation was that its p-value was lower than 0.05. Backward elimination of a candidate proxy signal happened when its p-value was larger than 0.1. The second method was to perform stepwise linear regressions with the candidate proxy signals recombined using principal component analysis (PCA): a technique that is known as principal component regression (PCR) (Hawkins, 1973). Because of the difference in the magnitude of the candidate proxy signals, they cannot be entered into the PCA directly. Therefore, data values were converted to z-scores (i.e. the difference between an individual value and the mean expressed in times the standard deviation) for each candidate proxy signal individually before entering the PCA. Next, the PCA was performed to recombine the candidate proxy signals using another built-in Matlab function. The first principal components that explain at least 95% of the variance in the dataset were then included as input parameters for the PCR. Criteria for principal components to enter or be removed from the regression equation were the same as described above.

The basic stepwise linear regression analyses and the PCR were performed twice: first with only the kinematic parameters as input variables and second with both the kinematic and kinetic variables. This was done because the collection and analysis of kinematics data is expected to be more feasible in a clinical context compared to the collection and analysis of kinetics data. Namely, the collection of sagittal plane kinematics would only require one video camera. The collection of sagittal plane kinetics would additionally require one or two force plates as well as spatial and temporal synchronisation between the force plates and the video camera. For each regression equation, the adjusted R^2 value and the p-value were described. Given the explorative nature of this study, these values will be purely described without passing judgement on the validity of the prediction made by these equations.

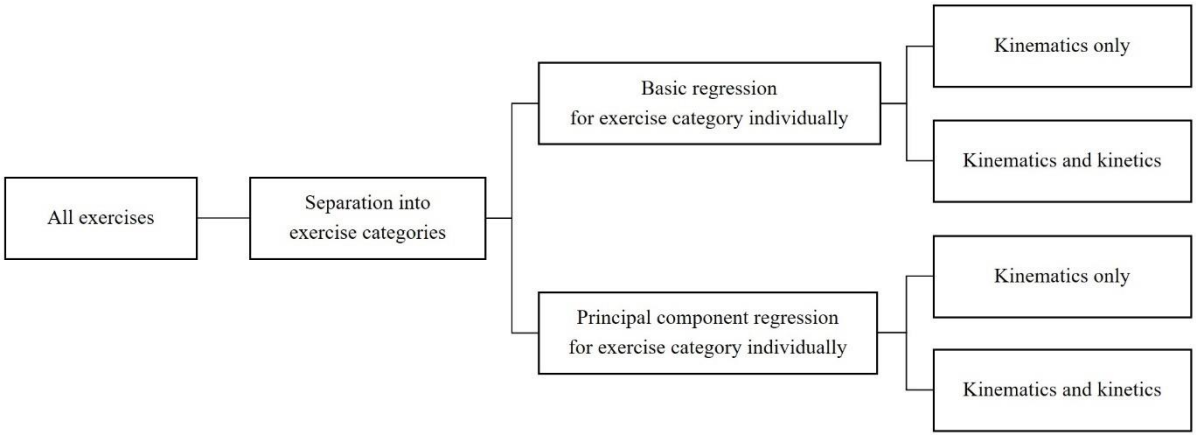

Figure 7.1. *Overview of the workflow of this exploratory analysis. First, the entire group of exercises was separated in exercise categories based on visual inspection of the scatter plots between the peak PTF and the proxy signals. Second, two methods of stepwise linear regression were performed per exercise category: basic regression with each candidate proxy signal as a separate independent variable and principal component regression. Each regression method was performed on the kinematic data alone and on a combination of kinematic and kinetic data.*

7.3 Results

7.3.1 Identification of relevant categories of exercises

Scatter plots of the peak PTF and all candidate proxy signals can be found in Supplementary analysis B. Visual inspection of these scatter plots revealed that the entire group of exercises can be separated into a group of fast exercises and a group of slow exercises based on the average velocities and the peak and average acceleration parameters (see Figure 7.2). The fast exercises typically overlapped with the jumping and hopping tasks, whereas the slow exercises included squats, lunges and step downs. Four exceptions were found for this categorisation: the forward lunge with step and the three non-knee specific exercises. Since the kinematic data of these four exercises are also rather different from one another and the number of data points for each exercise individually was low, they were removed from further analysis.

A second categorisation was performed within the fast exercises, i.e. the jumping and hopping tasks. Although all jumping and hopping exercises had high rectified average angular velocities, the exercises with only a push-off phase had low peak flexion velocities and the exercises with only a landing phase had low peak extension velocities. This is expected since the push-off exercises only include an extension movement and the landing exercises only a flexion movement. Based on these observations, the peak flexion velocities of the push-off exercises and the peak extension velocities of the landing exercises were excluded from the regression analyses. The plyometric exercises include both a flexion and extension movement, resulting in both high flexion and extension velocities. Therefore, no input parameters were excluded from the regression analyses for the plyometric exercises.

In the group of slow exercises, the number of input parameters was reduced before entering the regression analysis as well. More specifically, the peak extension angles were excluded because they occur during the upright positions of the squat, lunge and step down exercises. This position is known to be not related to the peak PTF since the peak PTF is reached around the point of peak flexion during these exercises.

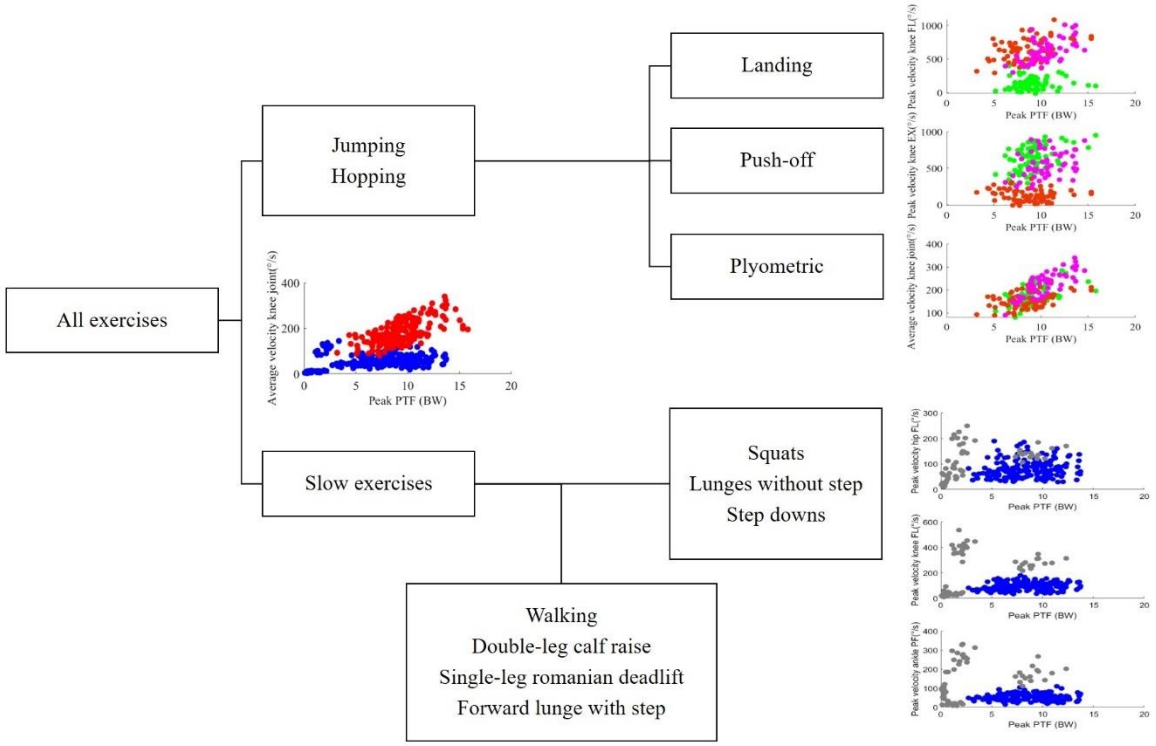

Figure 7.2. *Overview of the exercise categorisation process with exemplary scatter plots. First, the entire group of exercises was separated into a group of slow exercises (blue) and a group of fast exercises (red) based on the average velocities and the peak and average acceleration parameters. The fast exercises included the jumping and hopping tasks, whereas the slow exercises included the squats, lunges, step downs and non-knee specific exercises. Four outliers were found within the slow exercises based on the angular velocities and accelerations: the forward lunge with step and the three non-knee specific exercises (grey). These exercises were removed from further analysis. Within the jumping and hopping exercises, a separation between the landing (orange), push-off (green) and plyometric (purple) exercises. All jumping and hopping exercises had high rectified average angular velocities, but the exercises with only a push-off phase had low peak flexion velocities and the exercises with only a landing phase had low peak extension velocities. Plyometric exercises had both high peak flexion and extension velocities.*

PTF = patellar tendon force

BW = times body weight

FL = flexion

EX = extension

PF = plantar flexion

7.3.2 Regression analysis kinematics

For the regression analyses estimating the peak PTF based on the kinematic parameters as input variables, adjusted R^2 values were 0.398 ($p<0.001$) for the push-off exercises, 0.506 ($p<0.001$) for the squats, lunges and step downs, 0.575 ($p<0.001$) for the landing exercises and 0.663 ($p<0.001$) for the plyometric exercises (see Table 7.2). The push-off exercises required the largest number of input parameters i.e. four, while the landing and plyometric exercises required only two input parameters. The peak knee flexion angle and average knee flexion/extension velocity were the relevant input parameters across multiple exercise categories.

For the PCR based on the kinematic parameters, adjusted R^2 values ranged between 0.296 ($p<0.001$) for the push-off exercises and 0.641 ($p<0.001$) for the plyometric exercises. The adjusted R^2 values were always lower for the PCR compared to the regression with the kinematic parameters as direct input variables. Furthermore, all exercise categories except for the push-off exercises required a larger number of input parameters for the PCR: eight compared to three parameters for the squats, lunges and step downs and five compared to two parameters for the landing and plyometric exercises. The eigenvalues of each input parameter per principal component can be found in Supplementary analysis C.

Exercise category	Input parameters	Coefficient	Adjusted R^2 (p-value)
Squats/lunges/step downs	Intercept Average ankle velocity Peak knee FL angle Average knee velocity	-1.83 -0.140 0.120 0.0369	0.506 (<0.001)
Jumping/hopping push-off	Intercept Peak ankle PF velocity Peak ankle DF acceleration Average knee velocity Peak hip FL acceleration	0.873 0.0107 -0.0000641 0.0121 -0.000124	0.398 (<0.001)
Jumping/hopping landing	Intercept Peak knee FL angle Peak hip EX angle	3.46 0.0438 -0.0754	0.575 (<0.001)
Jumping/hopping plyometric	Intercept Peak ankle PF angle Average knee velocity	2.65 0.0434 0.0286	0.663 (<0.001)

Table 7.2.A. *Regression analysis for the estimation of peak PTF with kinematic parameters as input variables.*

Exercise category	Input parameters	Coefficient	Adjusted R^2 (p-value)
Squats/lunges/step downs	Intercept	8.52	0.434 (<0.001)
	PC1	0.104	
	PC2	0.392	
	PC3	0.626	
	PC4	-0.430	
	PC5	-0.368	
	PC6	0.618	
	PC7	0.345	
	PC9	-0.594	
Jumping/hopping push-off	Intercept	9.17	0.296 (<0.001)
	PC1	0.279	
	PC6	-0.422	
Jumping/hopping landing	Intercept	8.66	0.552 (<0.001)
	PC1	0.157	
	PC2	0.613	
	PC3	0.246	
	PC4	-0.440	
	PC6	-0.507	
Jumping/hopping plyometric	Intercept	10.5	0.641 (<0.001)
	PC1	0.358	
	PC2	0.208	
	PC3	0.371	
	PC5	-0.434	
	PC11	0.484	

Table 7.2.B. *Regression analysis for the estimation of peak PTF with principal components of the kinematic parameters as input variables.*

7.3.3 Regression analysis kinematics and kinetics

The regression analyses estimating the peak PTF based on kinematic and kinetic parameters can be found in Table 7.3. The adjusted R^2 values increased for most exercise categories – ranging from 0.717 ($p<0.001$) for squats, lunges and step downs to 0.824 ($p<0.001$) for push-off exercises – but not for the plyometric exercises. In fact, the regression equation for the plyometric exercises remained unaltered. In the other three exercise categories, the peak knee extensor moment was taken up as an input parameter and the peak hip extensor moment was included as well for the push-off exercises. Furthermore, new kinematic parameters were included in the regression equations for kinematics and kinetics compared to the regression equations based on kinematics alone.

Similar to the PCR based on the kinematic parameters, the PCR based on kinematic and kinetic parameters had lower adjusted R^2 values compared to the regression equations with the kinematic and kinetic parameters as direct input variables. The decrease in the adjusted R^2 values ranged between 0.032 for the landing exercises and 0.274 for the push-off exercises. Furthermore, the PCR required more input parameters for all exercise categories except for the push-off exercises.

Exercise category	Input parameters	Coefficient	Adjusted R^2 (p-value)
Squats/lunges/step downs	Intercept	-5.36	0.717 (<0.001)
	Peak ankle DF angle	-0.0404	
	Average ankle velocity	-0.168	
	Peak knee FL angle	0.110	
	Average knee velocity	0.0884	
	Peak knee EX moment	3.45	
	Average trunk acceleration	-0.0166	
Jumping/hopping push-off	Intercept	-9.66	0.824 (<0.001)
	Peak ankle DF angle	0.116	
	Peak ankle PF velocity	-0.00189	
	Average ankle velocity	0.0187	
	Peak knee FL angle	0.139	
	Peak knee EX moment	2.91	
	Peak hip FL angle	-0.0418	
	Peak hip EX moment	0.973	
	Peak trunk FL angle	-0.0418	
	Peak trunk EX velocity	0.00645	
Jumping/hopping landing	Intercept	-3.89	0.801 (<0.001)
	Average ankle velocity	0.141	
	Peak knee FL angle	0.0629	
	Peak knee EX angle	-0.0557	
	Peak knee EX moment	1.82	
	Peak hip EX angle	-0.0362	
Jumping/hopping plyometric	Intercept	2.65	0.663 (<0.001)
	Peak ankle PF angle	0.0434	
	Average knee velocity	0.0286	

Table 7.3.A. *Regression analysis for the estimation of peak PTF with kinematic and kinetic parameters as input variables.*

Exercise category	Input parameters	Coefficient	Adjusted R^2 (p-value)
Squats/lunges/step downs	Intercept	8.52	0.670 (<0.001)
	PC1	0.106	
	PC2	0.424	
	PC4	0.674	
	PC5	-0.287	
	PC6	0.382	
	PC7	0.671	
	PC8	0.216	
	PC9	0.830	
	PC10	1.16	
	PC12	0.332	
	PC14	0.863	
Jumping/hopping push-off	Intercept	9.17	0.550 (<0.001)
	PC1	0.277	
	PC6	-0.553	
	PC9	0.323	
	PC10	0.429	
	PC12	0.920	
Jumping/hopping landing	Intercept	8.66	0.769 (<0.001)
	PC1	0.176	
	PC2	0.565	
	PC3	0.144	
	PC4	-0.529	
	PC5	-0.311	
	PC6	-0.479	
	PC9	0.502	
	PC10	1.01	
	PC11	0.684	
	PC12	0.532	

Jumping/hopping plyometric	Intercept PC1 PC2 PC3 PC4 PC5	10.5 0.339 0.252 -0.286 0.271 -0.331	0.604 (<0.001)

Table 7.3.B. *Regression analysis for the estimation of peak PTF with principal components of the kinematic and kinetic parameters as input variables.*

7.4 Discussion

The objective of this study was to explore the potential for estimating peak PTF based on sagittal plane kinematics and kinetics across multiple rehabilitation exercises. The results of this study show that this potential is certainly present. Even the regression analyses with only kinematic input parameters showed promising results. The adjusted R^2 values of the basic regression analyses based on kinematics data alone ranged from 0.398 to 0.663. In other words, about half of the variations in the peak PTF can be explained by variations in the peak and average sagittal plane kinematic variables. These results show the potential for the development of regression equations based on sagittal plane kinematics, especially if the dataset would be expanded by including more rehabilitation exercises with a specific focus on kinematic variations. For example, in the current dataset important variations in the knee flexion angles were only present for the step down exercises, where the step height increased from 10 to 30 cm. The squats on the other hand were all performed up to a knee flexion angle of 90°. If a new data collection protocol would include different squat depths, the adjusted R^2 value of the regression equation would probably increase. A second example of the potential of sagittal plane kinematics, are the plyometric exercises. Here, the regression equation did not change when kinetics data were added as input parameters, meaning that the kinetics could not explain an additional portion of the variation in peak PTF compared to what had already been explained by the kinematics. If this dataset can be extended, covering a broader variation of movement patterns and jumping distances (Elvin et al., 2009; I. Janssen et al., 2015), the explained variation based on kinematics alone could even increase.

In the other three exercise categories (i.e. squats/lunges/step downs – push-off exercises – landing exercises), adding kinetic data to the regression equations did improve the adjusted R^2 value by 0.288 on average. These results show that high R^2 values can be obtained after a basic categorisation process, making each equation applicable to a variety of rehabilitation exercises. However, adding kinetics to the regression equation, comes at a cost. While recording sagittal plane kinematics would only require one video camera, sagittal plane kinetics would also require one or two synchronised force plates. This would make the data collection more expensive and less mobile. It would be valuable to know which of these kinetic parameters can be replaced by input parameters that are easily determined. For example, a separation can be made between single and double leg exercises, the mass of external weights can be included for squats, lunges and step downs or jump heights and distances can be included for the vertical and forward jump and hop exercises respectively (Elvin et al., 2009; I. Janssen et al., 2015).

The number of kinematic and kinetic input parameters investigated in this study was large. As expected, high levels of covariation between these input parameters were common (see Supplementary analysis D). If two input parameters would have a high correlation with the peak PTF but also high covariations, only one parameter would be included in the stepwise regression (Portney & Watkins, 2014a). For example, the peak hip flexion angle had a correlation of 0.30 or higher with the peak PTF in all exercise categories. However, this parameter was only included in one regression equation because it did not explain an additional percentage of the variation in peak PTF on top of more relevant parameters such as the peak knee flexion angle or the average knee velocity. In this search for parameters that can provide independent information on top of knee kinematics and/or kinetics, the ankle joint seems to be a region of interest. Seven out of the eight regression equations in this study included one or more ankle parameters. Furthermore, the peak ankle plantarflexion angle was one of the two parameters included in the regression equation of the plyometric exercises. A possible explanation for this observation might be the contribution of co-contractions from the bi-articular gastrocnemius muscle to the peak PTF. The suggestion that ankle kinematics play an important role in the magnitude of the PTF is further supported by Janssen et al. (2013), who included the average ankle dorsiflexion velocity as one of two kinematic parameters in their regression equation for the estimation of peak PTF during the landing of a lateral stop jump. The other kinematic parameter in their regression equation was the average trunk flexion velocity, whereas neither the peak trunk flexion velocity nor the rectified average trunk velocity was included in the regression equations of the landing tasks. Potential reasons for these different results include the difference in the candidate proxy signals selected in both studies and the different ways of calculating the peak PTF.

A potential way to overcome the barrier of high covariations among the input parameters, is to recombine these parameters using PCA. However, the results of this study show that the adjusted R^2 values were always smaller for the PCR compared to when the kinematic and kinetic parameters were used as direct input variables. Furthermore, PCR comes with some practical barriers to use in the field. First, each principal component is a recombination of all kinematic and kinetic parameters (see Supplementary analysis C). Therefore, all these parameters have to be measured and calculated before the PCR can be used. Second, the regression equations with kinematics and kinetics as direct input parameters are more straightforward to interpret compared to the principal component. In conclusion, PCR does not have an added value over basic stepwise linear regressions for the calculation of peak PTF across multiple rehabilitation exercises. However, PCR could still be beneficial for peak PTF

estimations within a specific rehabilitation exercise. Identifying proxy signals within a specific exercise has the advantage that PTF, kinematic and kinetic parameters have a similar curve over time. That would make it possible to decompose the PTF curve over time in its principal components, try to reconstruct each principal component from kinematic and/or kinetic parameters over time and determine the peak PTF from the reconstructed PTF curve (Ashall et al., 2023; Warmenhoven et al., 2021). Similarly, whole-body PCA may help to determine peak PTF values within specific rehabilitation exercises, since the identification of principal movements can resolve the barrier of high covariations among the input parameters (Daffertshofer et al., 2004; Federolf et al., 2014; Verheul et al., 2019).

This study was a secondary analysis of an existing dataset (study 4). The limitations of the current dataset and directions for the data collection of future studies have been discussed above. Also, the benefits and limitations of different stepwise linear regression approaches have been demonstrated in this study. Future research could use other techniques to predict peak PTF from kinematic and kinetic parameters. For example, the issue of high covariations between the candidate proxy signals could be overcome by a machine learning approach with artificial neural networks. While these techniques fall beyond the scope of the current exploratory analysis based on regression equations, results like the exercise categorisation are also relevant for researchers and engineers who attempt to predict peak PTF with a machine learning approach. A final limitation of the current study is the fact that the peak PTF values were calculated using musculoskeletal modelling. PTF values can only be directly measured in vivo through invasive techniques such as buckle transducers or optic fibre sensors (Fleming & Beynnon, 2004; Ravary et al., 2004; Roriz et al., 2014). Musculoskeletal modelling can be considered as a current best practice for non-invasive PTF measurements though, since it has been used in the past to calculate peak PTF over a large variety of physical activities (Garcia et al., 2022; Hagen, Van Rossom, et al., 2023; Hagen, Vanmechelen, et al., 2023; Lenhart et al., 2014; Mokhtarzadeh et al., 2014; Zellmer et al., 2019). However, we should still be aware that this method in itself comes with assumptions and estimation errors (discussed in detail in study 4 supplementary analysis C).

7.5 Conclusion

This study explored the potential of peak PTF estimation from sagittal plane kinematics and kinetics across several rehabilitation exercises. This explorative analysis identified four relevant categories of exercises and found several potential proxy signals per category. Furthermore, the use of PCR did not provide an added value over a basic stepwise linear regression. This information is an important first step for researchers and engineers who want to develop clinical measurement tools for PTF predictions in the future.

7.6 References

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Supplementary analysis A

Video of the rehabilitation exercises:

https://drive.google.com/file/d/1Nlby_v7CAonI_IlXDDA_cGLAi--DbmAa/view?usp=sharing

Supplementary analysis B

Scatter plots of the peak patellar tendon force and the kinematic and kinetic proxy signals: https://drive.google.com/drive/folders/1fPlSa--4sjuYpQu-_Dt-hfXYy93_4m3U?usp=sharing

Supplementary analysis C

Eigenvalues of each kinematic and/or kinetic parameter in the principal component regression: https://drive.google.com/drive/folders/1nIYDgAlTo8B63BIKb63I99XJ-4Vj7QeM?usp=sharing

Supplementary analysis D

Correlation matrix with the correlation coefficients between the peak patellar tendon force, the kinematic parameters and the kinetic parameters:
https://drive.google.com/drive/folders/18vuNJ52zLvFsYrxFFnCxz_cejHpu__2U?usp=sharing

Chapter 8

General discussion

8.1 Summary of main findings

The overall aim of this doctoral thesis was to improve the objective quantification of patellar tendon loading in a clinical practice setting. Throughout this thesis patellar tendon forces (PTF) and their progressions were objectively quantified during frequently used exercises in the rehabilitation of patellar tendinopathy. Furthermore, initial steps were taken towards measurement tools that can objectively quantify PTF in a clinical practice setting. This doctoral thesis showed that:

1. The effectiveness of step frequency alterations during running at different running speeds depends on the specific PTF parameter that is investigated. Step frequency alterations are equally effective at low and high running speeds for the peak PTF, less effective at high running speeds for the PTF impulse per kilometre and not effective for the peak rate of PTF development.
2. An external weight's mass and carrying position both influence the peak PTF during a forward lunge. The rate of increase in peak PTF with increasing mass is similar for all tested carrying positions: in one hand at the ipsilateral or contralateral side of the leading leg, or in two hands at the side or in front of the trunk.
3. The use of a decline board increases the peak PTF during a heavy weight single-leg squat. In contrast, increasing the mass of the external weight from 70% to 90% of the one repetition maximum does not result in a progressively higher peak PTF.
4. The progression of the peak PTF across a broad range of rehabilitation exercises are equivalent in patients with patellar tendinopathy and healthy individuals. Peak PTF progressions observed in healthy individuals can be applied to patients with patellar tendinopathy.
5. Physiotherapists have a good to excellent overall ability to estimate peak PTF during rehabilitation exercises compared to the current best practice musculoskeletal modelling approach. However, the load estimation can still be improved at the level of individual exercises since some high load exercises were underestimated and some low load exercises overestimated.
6. There is a potential for peak PTF estimations across multiple rehabilitation exercises from sagittal plane kinematics and kinetics and even from kinematics alone.

8.2 Implications for progressive tendon loading programs

8.2.1 An improvement on external load progressions

The objective of a mechanotherapeutic rehabilitation program for patellar tendinopathy is to gradually increase the load on the injured structure until habitual load levels can be tolerated again (Glasgow et al., 2015; Hart & Scott, 2012; Malliaras et al., 2015). Currently, load progressions throughout a rehabilitation program happen based on external instead of internal, patellar tendon specific load parameters. However, this approach is not optimal because of two reasons. First, the studies in this doctoral thesis clearly show that the increase in peak PTF is not necessarily proportional to the increase in external load (Glasgow et al., 2015; Soligard et al., 2016; Vanrenterghem et al., 2017). For example, the results of study 1 show that a 10% increase or decrease of the step frequency during running does not change the peak rate of PTF development and is less effective at high running speeds for the PTF impulse per kilometre. Furthermore, study 3 showed that the peak PTF does not increase while increasing the mass of the external weight from 70% to 90% of the one repetition maximum.

Second, changes in PTF parameters are not only influenced by external load parameters, but also by the movement pattern that individuals use to perform the exercise. For example, Zellmer et al. (2019) observed a larger peak patellar tendon stress – i.e. PTF divided by the cross-sectional area of the patellar tendon – when the knee was positioned in front of the toes compared to behind the toes. Similarly, Scattone Silva et al. (2017) observed a decreased peak PTF when a drop jump landing was performed with a more flexed trunk angle. These two studies have deliberately altered the movement execution to influence the peak PTF, but also natural variations in the movement pattern influence the peak PTF. This has been clearly shown in study 6 where variations in kinematic parameters could explain up to 66.3% of the variations in peak PTF during plyometric jumping and hopping tasks. Overall, study 6 has shown the potential for peak PTF estimations based on sagittal plane kinematics, measured with one video camera. Therefore, it is possible that in the future physiotherapists can evaluate the influence of variations in the movement pattern on PTF load parameters in a physiotherapy practice setting without the need for expensive equipment or advanced data analysis skills.

8.2.2 High peak patellar tendon forces do not require heavy weights

In the rehabilitation of patellar tendinopathy, heavy slow resistance training has been reported as the second of four stages within the clinical guidelines for progressive tendon loading (Malliaras et al., 2015). The concept of this stage is to gradually increase the peak PTF, while the rate of PTF development is kept low. Interestingly, the results of study 3 demonstrated that the peak PTF does not progressively increase when the mass of the external weight was increased from 70% to 90% of the one repetition maximum. Moreover, the results of this doctoral thesis showed that reaching high peak PTF values does not require the use of heavy external weights. For example, both studies 3 and 4 demonstrated that the use of a decline board is more effective for increasing the peak PTF than the use of external weights. Furthermore, the results of study 2 showed that changing the weight-carrying position can also increase the peak PTF during a forward lunge. Holding an external weight of 10% body mass in the ipsilateral hand resulted in mean peak PTF values that corresponded to an external weight of 30% body mass in the contralateral hand. Not needing such heavy external weights to reach high peak PTF values paves the way for low-cost, home-based resistance training programs for the rehabilitation of patellar tendinopathy (Breda et al., 2021).

A few remarks should be made though in favour of strength training with heavy weights. First, the supplementary analysis of the heavy weight single-leg squats (study 3) showed that the hip rather than the knee extensors produced the additional joint moments to lift the increasingly heavier weights. As mentioned in the general introduction, one goal of the rehabilitation program is to keep the load on the surrounding uninjured structures as habitual as possible or even increase it (Cook & Docking, 2015; Malliaras et al., 2015; Powers, 2010; Silva et al., 2015). Second, heavy weight exercises should also be discussed in terms of pain reduction. Rio et al. (2015) discovered that isometric exercises at 70% of the one repetition maximum triggered an analgesic response in the central nervous system. Although the superiority of isometric over isotonic exercises is still a topic of discussion, heavy loads seem a necessity for this neurologically induced pain reduction (Holden et al., 2019; Rio et al., 2015, 2017; van Ark et al., 2016). The question that can now be asked, is whether a high peak PTF alone is sufficient to provoke this analgesic response. For example, does a single-leg squat at 70% of the one repetition maximum generate the same amount of pain reduction as the single-leg squat at 90% of the one repetition maximum?

8.2.3 An improvement on pain-based load progressions

In the current clinical practice setting, internal patellar tendon load progressions are monitored through pain. The physiotherapeutic guidelines suggest pain scores up to three on a ten-point numeric rating scale during the exercise program and pain scores on a loading test returning to its baseline value within 24 hours after the end of the exercise program (Escriche-Escuder et al., 2020; Jonsson & Alfredson, 2005; Malliaras et al., 2015). The objective of this pain-based loading test is to determine whether or not the patellar tendon load during the rehabilitation program was sufficiently high to improve the patellar tendon load capacity. An important downside of pain-based load progressions, is its limited applicability in patellar tendinopathy patients with a dominant psychosocial pain profile (Hanlon et al., 2021). In this patient group – characterised by high levels of kinesiophobia and pain catastrophising – there is a partial or complete uncoupling between pain on the one hand and structural and functional aspects on the other hand (Edgar et al., 2022; Hanlon et al., 2021). Fortunately, the results of this doctoral thesis also provide some clinical tools for this patient group. In the current absence of clinical PTF measurement tools, the results of study 5 provide evidence that patients with patellar tendinopathy can be confident in the load estimating capacity of their physiotherapist. Furthermore, the objectively quantified peak PTF progressions in this doctoral thesis can be used for patient education.

8.3 Methodological considerations

An important methodological aspect of every study in this doctoral thesis, is advanced musculoskeletal modelling (Delp et al., 2007; Seth et al., 2018). This advanced musculoskeletal modelling approach provides important advantages for the calculation of PTF over the division of the knee extensor moment by the angle-specific patellar tendon moment arm (Herzog & Read, 1993) since it can take the co-contractions from antagonistic muscles into account. It should be noticed though that the advanced musculoskeletal modelling approach has some assumptions and simplifications as well. One important assumption that has been discussed, was the choice of the patellar tendon moment arm. Supplementary analysis C of study 4 clearly indicated a discrepancy between the calculated muscle activations from static optimisation and the measured muscle activations from electromyography. Upon further investigation, a clear indication was given that the length of the patellar tendon moment arm could explain this discrepancy. More specifically, evidence was given that the patellar tendon moment arm of the musculoskeletal model was too small in more flexed knee angles. This resulted in an angle-specific overestimation of the peak PTF, with larger overestimation in more flexed knee joint

angles. However, irrefutable evidence in this case can only come from measuring the patellar tendon moment arm with dynamic medical imaging, which was not available for this study. Nevertheless, it should be discussed how a change in the patellar tendon moment arm of the model would influence the results of the different studies in this doctoral thesis. For study 1-3, the results would only be minimally affected. For each of these studies, variations of the same exercise – with the same knee flexion angles in each variation – were investigated. Furthermore, each study had a repeated measures design where each participant is compared against themself. Hence, the conclusions about progressions in PTF parameters in response to the investigated exercise variations are not influenced by changes in the patellar tendon moment arm, even if the absolute PTF values might change. In study 4, the correlation between the peak PTF values of patients with patellar tendinopathy and the healthy individuals was not influenced by changes in the patellar tendon moment arm since both groups performed the exercises with the same knee angles. However, supplementary analysis C of study 4 clearly showed that the potential impact of changes in the patellar tendon moment arm on the mean values, standard deviations and rankings of the rehabilitation exercises for peak PTF, is severe. Since the results of study 5 and 6 depend on these absolute peak PTF values, these two studies are the most impacted by changes in the patellar tendon moment arm. From study 5, we can conclude that physiotherapists have a good to excellent overall ability to estimate peak PTF compared to the current best practice musculoskeletal modelling approach, but future studies should verify these results. Nevertheless, it can still be concluded that the PTF estimating capacity of the subgroups based on years of clinical experience and specialisations is not significantly different. Considering study 6, the candidate proxy signals identified in this study can change when the patellar tendon moment arm is altered for the calculation of the peak PTF values across exercises. However, the description of the methodology how proxy signals for peak PTF can be identified across multiple rehabilitation exercises and the suggestions for future data collection and analysis procedures in the discussion, is still relevant for researchers and engineers who want to develop clinical measurement tools for PTF estimations in the future.

Determining the accuracy of a measurement tool for the objective quantification of PTF progressions is one aspect. However, another important aspect in the validation process, is to determine the minimal clinically important difference (Portney & Watkins, 2014b). When this minimal clinically important difference is not known – which is the case for PTF progressions at this moment – it is also difficult to assess what the minimal accuracy of a measurement tool should be. Take for example the regression equations from study 6, describing PTF from proxy signals. Without knowing the minimal clinically important difference, it is hard to establish

how accurate measurement tools based on these proxy signals should be in order to become useful in a clinical practice setting or in clinical scientific research. This is also the case for the subjective peak PTF estimates by physiotherapists (study 5). In the absence of a minimal clinically important difference, the cut-off value for overestimation or underestimation in this study were arbitrarily set to a medium Cohen's d effect size of 0.5. However, it is possible that this arbitrary criterion was actually not strict enough or too strict. Therefore, identifying this minimal clinically important difference is an important objective for future research. In other words, future research needs to determine how much PTF progression is minimally necessary to trigger positive adaptations.

The minimal clinically important difference should not only be identified for the peak PTF, which was the parameter of interest for most studies in this doctoral thesis. As mentioned in the general introduction of this thesis, resultant patellar tendon load is not only determined by the peak PTF. Due to mechanical phenomena like creep, stress-relaxation and hysteresis, the stress-strain relationship of the patellar tendon is influenced by the contraction velocity and the time under tension (Kelc et al., 2013; Kösters et al., 2014; Pearson et al., 2007). Hence, the rate of PTF development and the PTF impulse are also important load parameters (Benson et al., 2021; Firminger et al., 2020; Tayfur et al., 2022). Since standardized instructions or structured variations in the time under tension or the execution velocity were only given in study 1, this was the only study where calculations of the rate of PTF development and the PTF impulse were deemed appropriate. Future research could objectively quantify the rate of PTF development and the PTF impulse for other rehabilitation exercises as well. Furthermore, in the context of determining the minimal clinically important difference for PTF parameters, it should be determined what the contribution of the peak PTF, the rate of PTF development and the PTF impulse to the resultant patellar tendon load and damage is (W. B. Edwards, 2018; Scattone Silva et al., 2023).

8.4 Directions for future research

This doctoral thesis has taken some important steps towards the optimisation of patellar tendon loading in musculoskeletal rehabilitation. However, the results of this thesis inspire additional research questions to optimise patellar tendon loading even further. In their turn, these additional research questions can be investigated in future research. Some of the suggestions for future research have been indirectly presented in earlier sections of this discussion, yet it is worthwhile to explicitly summarise the most important suggestions here below.

A first important step towards further optimisation of patellar tendon loading in musculoskeletal rehabilitation, is gaining a continued further understanding of PTF progressions through a combination of direct in vivo measurements and advanced musculoskeletal modelling. The need for further elaboration of the currently used musculoskeletal model has been clearly demonstrated in this doctoral thesis. Along the same line, it would be interesting to build up a dataset of rehabilitation exercises where the PTF is directly measured. This dataset can further strengthen the validation of new musculoskeletal modelling approaches for the calculation of PTF. Unfortunately, the direct measurement of PTF in vivo currently requires the surgical implantation of a buckle transducer or optic fibre sensor (Fleming & Beynnon, 2004; Ravary et al., 2004; Roriz et al., 2014). However, the shear wave tensiometer might provide a non-invasive alternative in the future (Martin et al., 2018). This device measures the tensile patellar tendon stress during exercises by measuring the propagation speed of a shear wave along the tendon. Since the propagation speed for a given tensile stress is tissue-specific, it is not possible to covert this propagation speed directly to a PTF value (Crawford et al., 2023). However, an increase in the propagation speed can be an indirect validation for PTF progressions during exercise variations.

In a second step towards the optimisation of patellar tendon loading, prospective observational studies need to determine the pace at which the different PTF parameters need to progress throughout the rehabilitation program, i.e. determine the minimal clinically important difference of each PTF parameter to trigger positive adaptations. Three forms of adaptation should be monitored. First, changes in the patellar tendon pain can be monitored through pain-based loading tests or through the VISA-P questionnaire (Escriche-Escuder et al., 2020; Johannes Zwerver et al., 2009). Second, structural changes in the patellar tendon can be monitored through medical imaging techniques such as shear wave elastography or ultrasonic tissue characterisation (Rabello et al., 2021; Taljanovic et al., 2017). Finally, functional adaptations can be monitored though functional tests such as quadriceps muscle strength testing, maximum jump heights and distances or agility T sprint tests.

In a final step towards the optimisation of patellar tendon loading, the clinical utility of normative PTF dataset and/or the clinical PTF measurement tools needs to be validated against the current best practice: PTF estimations by physiotherapists. This would require a randomised controlled trial with two intervention arms: one in which the physiotherapist is assisted by a normative PTF dataset or clinical PTF measurement tool and one in which the physiotherapist depends on his/her subjective estimations alone. Both the content of the rehabilitation programs and the resultant effects on pain reduction, structural adaptations and functional improvements can then be compared between both interventions.

8.5 Conclusion

Through this doctoral thesis, musculoskeletal rehabilitation has taken some important steps in the optimisation of patellar tendon loading. This thesis has objectively quantified PTF progressions over several exercise variations, overthrowing some common assumptions in the process. Moreover, evidence has been provided that PTF progressions across a broad range of rehabilitation exercises are equivalent in patients with patellar tendinopathy and healthy individuals. Furthermore, this thesis determined the PTF estimating accuracy of physiotherapists and took initial steps towards objective PTF measurements in a clinical practice setting. In result, this thesis has provided important advancements in knowledge and tools for physiotherapists to create their rehabilitation programs. This will allow physiotherapists and their patients with patellar tendinopathy to exercise with more confidence.

8.6 References

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Summary

Mechanotherapy is an important concept for physiotherapy in general and for the rehabilitation of patellar tendinopathy specifically. During a mechanotherapeutic rehabilitation program, the load on the injured structure is gradually increased according to the principles of supraphysiological overload and supercompensation until habitual load levels can be tolerated again. At present, progressive patellar tendon loading is monitored through external load parameters such as the mass of an external weight during strength training. However, the same external load can lead to different internal loads on specific structures. Internal patellar tendon load progressions during the rehabilitation of patellar tendinopathy are currently monitored through a combination of visual inspection, clinical reasoning and pain indications. Rehabilitation programs based on objectively quantified patellar tendon load progressions would offer an important improvement from a mechanotherapeutic perspective. Unfortunately, the objective quantification of patellar tendon forces requires expensive, specialised and time-consuming techniques such as advanced musculoskeletal modelling. This makes the objective quantification of patellar tendon forces in a clinical practice setting currently impossible.

The overall objective of this doctoral thesis was to improve the objective quantification of patellar tendon force progressions in a clinical practice setting. Since only three studies have objectively quantified patellar tendon forces through advanced musculoskeletal modelling so far, the first specific objective of this thesis was to further objectively quantify patellar tendon forces during frequently used rehabilitation exercises for patellar tendinopathy. Studies 1-3 objectively quantified patellar tendon forces during running, lunges and squats respectively. More specifically, the (interacting) effects of changes in multiple external load parameters were investigated. During running (study 1), the effectiveness of step frequency alterations at different running speeds was investigated. The results of this study showed that step frequency alterations are equally effective at low and high running speeds for the peak patellar tendon force, less effective at high running speeds for the patellar tendon force impulse per kilometre and not effective for the peak rate of patellar tendon force development. During forward lunges (study 2), the effects of an external weight's mass and carrying position on the peak patellar tendon force were investigated. Here, the results showed that the external weight's mass and carrying position both influence the peak patellar tendon force. Increases of peak patellar tendon force with increasing mass were similar for all positions. Finally, the effects of an external weight's mass and the use of a decline board on the peak patellar tendon force were investigated during heavy load single-leg squats (study 3). Interestingly, only the use of a decline board increased the peak patellar tendon force. Increasing the mass of the external weight from 70%

to 90% of the one repetition maximum did not result in a progressively higher peak patellar tendon force.

Since studies 1-3 have been conducted with healthy participants, the question can be asked whether patients with patellar tendinopathy may have different patellar tendon force progressions between exercises. The exercise order to achieve a gradual patellar tendon force progression may well be different in patients with patellar tendinopathy, for example because of pain-avoidance strategies. The objective of study 4 was to assess to which extent the peak patellar tendon force progressions for a large number of rehabilitation exercises are equivalent in patients with patellar tendinopathy and healthy individuals. Seven patients with patellar tendinopathy and seven healthy individuals performed 28 rehabilitation exercises. The patients with patellar tendinopathy executed the rehabilitation exercises with mild pain scores, while the healthy individuals performed the exercises pain-free. Nevertheless, the peak patellar tendon force progressions in both groups showed near-perfect equivalence. It can therefore be concluded that objectively quantified peak patellar tendon force progression principles for healthy individuals can be transferred to patients with patellar tendinopathy.

This doctoral thesis had the overall objective to improve the objective quantification of patellar tendon force progressions in a clinical practice setting. Within this context, it was important to establish the accuracy level of patellar tendon force estimations made by physiotherapists based on their visual inspection and clinical reasoning skills. In study 5 a video-based survey was sent to 50 actively practicing physiotherapists and 50 final year master students. These physiotherapists estimated the peak patellar tendon force on a 100-point numeric rating scale for one individual performing 19 rehabilitation exercises. The results were compared against objectively quantified peak patellar tendon forces based on advanced musculoskeletal modelling. The results showed that the overall ability to estimate peak patellar tendon forces was good to excellent for the entire group and all subgroups. However, with six overestimated and four underestimated exercises, load estimations can still be improved at the level of individual exercises.

In study 6 an initial step was then taken towards measurement tools that can help objectively quantify patellar tendon forces in a clinical practice setting. This was done by identifying potential kinematic and kinetic proxy signals for peak patellar tendon forces over multiple rehabilitation exercises. Based on basic stepwise linear regression and principal component regression, it was demonstrated that there is a potential for peak patellar tendon force estimations from sagittal plane kinematics and kinetics and even from kinematics alone. This is good news for physiotherapists since the collection of sagittal plane kinematic data would only require one video-camera.

In conclusion, this doctoral thesis has taken some important steps towards improved progressive patellar tendon loading programs. The thesis has taken initial steps towards the development of clinical measurement tools for the objective quantification of patellar tendon forces. While these clinical measurement tools are further developed, this thesis has provided evidence that patients with patellar tendinopathy can already now be confident in the progressive exercise programs created by physiotherapists. Especially now that physiotherapists have access to patellar tendon force progressions – objectively quantified through advanced musculoskeletal modelling – over several exercise variations. Physiotherapists can use these objectively quantified patellar tendon force progressions to improve their patellar tendinopathy rehabilitation programs even further.

Samenvatting

Mechanotherapie is een belangrijk concept voor de kinesitherapie in het algemeen en voor de revalidatie van patellapeestendinopatie in het bijzonder. Tijdens een mechanotherapeutisch revalidatieprogramma wordt de belasting op de geblesseerde structuur gradueel opgedreven volgens de principes van suprafysiologische overbelasting en supercompensatie totdat het gebruikelijke belastingniveau opnieuw getolereerd kan worden. Momenteel wordt de progressiviteit van de patellapeesbelasting opgevolgd via externe belastingparameters zoals de massa van een extern gewicht tijdens krachttraining. Het probleem is echter dat eenzelfde externe belasting kan leiden tot verschillende interne belastingen op verschillende specifieke structuren. Op dit moment worden de interne patellapeesbelasting tijdens de revalidatie van patellapeestendinopathie opgevolgd door een combinatie van visuele inspectie, klinisch redeneren en pijnindicaties. Revalidatieprogramma's gebaseerd op objectief gekwantificeerde patellapeesbelastingen zouden dan ook een belangrijke vooruitgang zijn vanuit een mechanotherapeutisch standpunt. Spijtig genoeg vereist de objectieve kwantificatie van patellapeeskrachten dure, gespecialiseerde en tijdrovende technieken zoals geavanceerd musculoskeletaal modeleren. Dit maakt de objectieve kwantificatie van patellapeeskrachten in een klinische praktijk momenteel onmogelijk.

De algemene doelstelling van deze doctoraatsthesis was om de objectieve kwantificatie van progressies in patellapeeskrachten te verbeteren in de klinische praktijk. Aangezien tot nog toe slechts drie studies patellapeeskrachten objectief gekwantificeerd hebben met geavanceerd musculoskeletaal modeleren, was de eerste specifieke doelstelling van deze thesis om patellapeeskrachten verder objectief te kwantificeren tijdens veelgebruikte revalidatieoefeningen voor patellapeestendinopathie. Studies 1-3 hebben objectief patellapeeskrachten gekwantificeerd tijdens respectievelijk lopen, uitvalpassen en squats. Meer specifiek werden de (interagerende) effecten van veranderingen in meerdere externe belastingparameters onderzocht. Tijdens het lopen (studie 1) werd de effectiviteit van aanpassingen in de stapfrequentie op verschillende loopsnelheden onderzocht. De resultaten van deze studie toonden aan dat de aanpassingen in de stapfrequentie even effectief zijn bij hoge en lage loopsnelheden voor de piek patellapeeskracht, minder effectief zijn bij hoge dan bij lage loopsnelheden voor de patellapeeskracht impuls per kilometer en niet effectief zijn voor de snelheid van patellapeeskracht ontwikkeling. Tijdens de voorwaartse uitvalpas (studie 2) werden de effecten van de massa en positie van een extern gewicht op de piek patellapeeskracht onderzocht. Hier toonden de resultaten dat zowel de massa als de positie van het extern gewicht de piek patellapeekracht beïnvloedden. Bovendien nam de piek patellapeeskracht bij een stijgende externe massa met eenzelfde hoeveelheid toe in alle getest posities. Tot slot werden

de effecten van een zwaar extern gewicht met toenemende massa en het gebruik van een hellend vlak op de piek patellapeeskracht onderzocht tijdens een eenbenige squat (studie 3). Vreemd genoeg zorgde enkel het hellend vlak voor een toename in de piek patellapeeskracht. De massa van het extern gewicht verhogen van 70% naar 90% van het maximale gewicht resulteerde niet in een progressieve toename van de piek patellapeeskracht.

Aangezien studies 1-3 uitgevoerd werden met gezonde deelnemers, kan men zich de vraag stellen of patiënten met patellapeestendinopathie een verschillend progressiepatroon tussen oefeningen zouden vertonen. De volgorde van oefeningen om een graduele toename van de patellapeeskracht te bekomen kan namelijk verschillend zijn voor patiënten ten opzichte van gezonde personen omwille van bijvoorbeeld pijnvermijdende bewegingsstrategieën. De doelstelling van studie 4 was daarom te bepalen tot op welke hoogte de piek patellapeeskrachten gelijkmatig toenemen bij patiënten met patellapeestendinopathie en gezonder personen over een brede waaier aan revalidatieoefeningen. Zeven patiënten met patellapeestendinopathie en zeven gezonde personen voerden een totaal van 28 revalidatieoefeningen uit. De patiënten voerden deze oefeningen uit met milde pijnscores terwijl de gezonde personen de oefeningen pijnvrij uitvoerden. Desalniettemin nam de piek patellapeeskracht quasi-perfect gelijkmatig toe in beide groepen. Daardoor kunnen we concluderen dat de objectief gekwantificeerde progressies in de patellapeeskrachten bij gezonde personen eveneens toegepast kunnen worden bij patiënten met patellapeestendinopathie.

Deze doctoraatsthesis had als algemene doelstelling om de objectieve kwantificatie van progressies in patellapeeskrachten te verbeteren in de klinische praktijk. In deze context was het belangrijk om te bepalen hoe accuraat kinesitherapeuten patellapeeskrachten kunnen inschatten op basis van hun visuele inspectie- en klinisch redeneervermogen. In studie 5 werd een vragenlijst met video's opgestuurd naar 50 actief werkende kinesitherapeuten en 50 laatstejaarsstudenten uit de master kinesitherapie. Zij schatten de piek patellapeeskracht in van een persoon die 19 revalidatieoefeningen uitvoerde op een 100-punten numerieke schaal. De resultaten werden vergeleken met de objectief gekwantificeerde piek patellapeeskrachten op basis van geavanceerd musculoskeletaal modelleren. De resultaten toonden aan dat de algemene inschattingscapaciteit goed tot uitstekend was voor de gehele groep en voor alle subgroepen. Desondanks kan het inschattingsvermogen op het niveau van individuele oefeningen nog wel verbeterd worden, aangezien zes oefeningen overschat en vier oefeningen onderschat werden.

In studie 6 tot slot, werden initiële stappen gezet voor de ontwikkeling van klinische meettoestellen die kunnen helpen patellapeeskrachten objectief te kwantificeren in de klinische praktijk. Dit werd gegaan door potentiële kinematische en kinetische proxy signalen voor de piek patellapeeskracht te identificeren voor meerdere revalidatieoefeningen tegelijk. Op basis van stapsgewijze lineare regressie en principal component regressie, werd aangetoond dat er potentieel is voor het inschatten van piek patellapeeskrachten op basis van kinematische en kinetische parameters uit het sagittale vlak en zelfs op basis van kinematische parameters alleen. Dit is goed nieuws voor kinesitherapeuten aangezien er voor het meten van kinematische parameters in het sagittale slechts een videocamera nodig is.

Samenvattend kunnen we stellen dat deze doctoraatsthesis enkele belangrijke stappen heeft gezet richting verbeterde oefenprogramma's met een progressieve patellapeesbelasting. De thesis heeft initiële stappen gezet in de ontwikkeling van klinische meetinstrumenten die patellapeeskrachten objectief kunnen kwantificeren. Terwijl deze klinische meetinstrumenten verder ontwikkeld worden, heeft deze thesis aangetoond dat patiënten met patellapeestendinopathie ook nu al vertrouwen kunnen hebben in de oefenprogramma's die door hun kinesitherapeuten. Zeker nu kinesitherapeuten toegang hebben tot patellapeeskrachten – objectief gekwantificeerd via geavanceerd musculoskeletaal modelleren – voor verschillende oefenvarianten. Kinesitherapeuten kunnen deze objectief gekwantificeerde patellapeeskrachten gebruiken om hun oefenprogramma's voor patiënten met patellapeestendinopathie nog verder te verbeteren.

Résumé

La mécanothérapie est un concept important pour la physiothérapie en général et pour la réadaptation de la tendinopathie patellaire en particulier. Pendant un programme de réadaptation mécanothérapeutique, la charge sur la structure blessée est progressivement augmentée selon les principes de la surcharge supraphysiologique et la surcompensation jusqu'à les charges habituelles sont de nouveau tolérables. À présent, la progressivité de la charge au tendon patellaire est surveillée par des paramètres de charge externe comme la masse d'un poids externe pendant un entraînement en force. Le problème est que la même charge externe peut mener aux différentes charges internes sur des différentes structures spécifiques. À ce moment-ci, la charge interne sur le tendon patellaire est surveillée pendant un programme de réadaptation par une combinaison de l'inspection visuelle, du raisonnement clinique et des indications de douleur. Des programmes de réadaptation à base des charges sur le tendon patellaire qui sont objectivement quantifiés, poseront une amélioration importante de la perspective mécanothérapeutique. Malheureusement, la quantification objective des forces sur le tendon patellaire demande des techniques chères et spécialisées qui prennent beaucoup de temps, comme la modélisation musculosquelettique avancée. Par conséquent, la quantification objective des forces sur le tendon patellaire n'est pas possible dans le contexte d'un cabinet clinique.

L'objectif général de cette thèse doctorale était l'amélioration de la quantification objective des progressions des forces sur le tendon patellaire dans le contexte d'un cabinet clinique. Parce que seulement trois études scientifiques ont objectivement quantifié des forces sur le tendon patellaire avec une modélisation musculosquelettique avancée, le premier objectif spécifique de cette thèse était de continuer la quantification objective des forces sur le tendon patellaire pendant des exercices de réadaptation. Pour les études scientifiques 1-3, les forces sur le tendon patellaire ont été objectivement quantifiées pour : la course à pied, les fentes et les squats. Plus précisément, les effets (interagissants) des modifications de plusieurs charges externes ont été déterminés. Pendant la course à pied (étude 1), l'efficacité des modifications de la fréquence des pas aux différentes vitesses a été étudiée. Les résultats ont montré que les modifications de la fréquence des pas sont aussi efficaces aux vitesses lentes et rapides pour la force maximale du tendon patellaire, moins efficaces aux vitesses rapides pour l'impulsion de force du tendon patellaire par kilomètre et pas efficaces pour la vitesse de développement de force du tendon patellaire maximale. Pendant les fentes (étude 2), les effets de la masse et de la position d'un poids externe sur la force maximale du tendon patellaire ont été étudiés. Les résultats ont montré que la masse et la position influencent la force maximale du tendon patellaire. Les augmentations de la force maximale du tendon patellaire avec une augmentation de la masse

étaient similaires pour toutes les positions. Enfin, les effets de la masse d'un poids externe lourd et de l'utilisation d'une planche déclinante sur la force maximale du tendon patellaire ont été étudiés pendant des squats unipodales (étude 3). Il est intéressant d'observer que seulement l'utilisation d'une planche déclinante a augmenté la force maximale du tendon patellaire. Par contre, l'augmentation de la masse du poids externe de 70% à 90% de la résistance maximale du poids externe n'a pas augmenté la force maximale du tendon patellaire.

Vu que des personnes en bonne santé ont participé aux études 1-3, on peut se demander si des patients avec une tendinopathie patellaire ont des progressions différentes de la force maximale du tendon patellaire. L'ordre des exercices de rééducation pour obtenir une progression graduelle de la force maximale du tendon patellaire peut être différent pour des patients avec une tendinopathie à cause des stratégies de mouvement pour éviter la douleur. L'objectif de l'étude 4 était donc de déterminer le degré d'équivalence des progressions de la force maximale du tendon patellaire dans un groupe de patients avec une tendinopathie patellaire et un groupe de personnes en bonne santé sur une grande variété des exercices de rééducation. Sept patients avec une tendinopathie patellaire et sept personnes en bonne santé ont effectué 28 exercices de rééducation. Les patients ont effectué ces exercices avec une douleur légère et les personnes en bonne santé ont effectué ces exercices sans douleur. Malgré la douleur, l'équivalence des progressions de la force maximale du tendon patellaire était quasiment parfaite. On peut donc conclure que les principes de progression de la force maximale du tendon patellaire observés chez des personnes en bonne santé peuvent être transférés aux patients avec une tendinopathie patellaire.

L'objectif général de cette thèse doctorale était l'amélioration de la quantification objective des progressions des forces sur le tendon patellaire dans le contexte d'un cabinet clinique. Dans ce contexte, il était important d'établir le niveau de précision des estimations de la force du tendon patellaire par des physiothérapeutes basées sur l'inspection visuelle et le raisonnement clinique. Dans l'étude 5, une enquête vidéo a été envoyée à 50 physiothérapeutes en activité et 50 étudiants qui sont dans leur dernière année de master en physiothérapie. Ces participants ont estimé la force maximale du tendon patellaire pour une personne qui a effectué 19 exercices de rééducation. Les résultats ont montré que la précision globale des estimations était bonne et même excellente pour le groupe complet et tous les sous-groupes. Cependant, avec six exercices surestimés et quatre exercices sous-estimés, la précision des estimations peut encore améliorer au niveau des exercices individuels.

Dans l'étude 6, une première étape a été prise dans le développement des outils de mesure qui peuvent aider à quantifier objectivement les forces du tendon patellaire dans une pratique clinque. Cela a été fait en identifiant des signaux cinématiques et cinétiques potentiels qui sont utiles pour l'estimation de la force maximale du tendon patellaire pour multiples exercices de rééducation. À base d'une régression linéaire par paliers et d'une régression des composantes principales, il a été démontré qu'il y a un potentiel pour l'estimation la force maximale du tendon patellaire avec des signaux cinématiques et cinétiques du plan sagittal et même avec des signaux cinématiques seuls. C'est une bonne nouvelle pour les physiothérapeutes car on n'a besoin que d'une seule caméra vidéo pour collecter des signaux cinématiques du plan sagittal.

En conclusion, cette thèse doctorale a pris quelques étapes importantes vers l'amélioration des programmes de réadaptation pour la tendinopathie patellaire. La thèse a fait les premiers pas vers le développement d'outils de mesure qui peuvent quantifier objectivement les forces du tendon patellaire dans une pratique clinque. Pendant que ses outils de mesure continuent à se développer, cette thèse a prouvé que des patients avec une tendinopathie patellaire peuvent déjà avoir confiance en les programmes de réadaptation créés par des physiothérapeutes. Surtout maintenant que les physiothérapeutes ont accès aux forces du tendon patellaire – objectivement quantifiées avec une modélisation musculosquelettique avancée – pour plusieurs variations d'exercices. Les physiothérapeutes peuvent utiliser ces progressions de la force du tendon patellaire objectivement quantifiées pour améliorer leurs programmes de réadaptation encore plus.

Appositions

In the world of rehabilitation, you witness the beauty of Darwin's evolution theory on a microscopic scale. Injured people survive and improve their quality of life through their inherent adaptive capabilities.

If we want to tackle the great challenges of our time, we are going to need every single bit of talent that we have. Within this world of nearly eight billion people, each individual contribution counts and should be recognised.

Language barriers pose a major limitation to implement clinical research around the globe, especially in developing countries. It is therefore high time that scientific papers can be submitted and read in multiple languages.

In de wereld van revalidatie, ben je getuige van de schoonheid van Darwins evolutietheorie op microscopische schaal. Geblesseerde personen overleven en verbeteren hun levenskwaliteit door hun inherente aanpassingsvermogen.

Als we de grote uitdagingen van onze tijd willen aanpakken, zullen we elk beetje talent kunnen gebruiken. In deze wereld van bijna acht miljard mensen, is elke individuele bijdrage belangrijk en zou die erkent moeten worden.

Taalbarrières vormen een belangrijke beperking voor de implementatie van klinisch onderzoek wereldwijd, en zeker in ontwikkelingslanden. Het is daarom hoog tijd dat wetenschappelijke artikels ingediend en gelezen kunnen worden in meerdere talen.

Dans le monde de la réadaptation, on témoigne la beauté de la théorie d'évolution de Darwin à une échelle microscopique. Des personnes blessées survivent et améliorent leur qualité de vie par leurs capacités adaptives inhérentes.

Si nous voulons faire face aux grands défis de notre époque, nous aurons besoin de tous les talents que nous disposons. Dans ce monde de presque huit milliards d'habitants, chaque contribution individuelle compte et doit être reconnue.

Des barrières linguistiques sont un obstacle majeur pour l'implémentation de la recherche clinique au monde, particulièrement aux pays en voie de développement. Il est donc urgent que des articles scientifiques peuvent être soumis et lus en plusieurs langues.

Professional career

Professional career

Michiel Hagen was born on the 26th of May 1996 in Leuven, Belgium. In 2014, he graduated from high school at Sint-Clara College, Arendonk, Belgium. Magna cum laude graduations as Bachelor and Master in Rehabilitation Sciences and Physiotherapy at KU Leuven, Belgium followed in 2017 and 2019 respectively. Immediately after his Master graduation, he started as a doctoral researcher at the Musculoskeletal Rehabilitation Research Group of the Department of Rehabilitation Sciences, KU Leuven, Belgium under supervision of Prof. Dr. Jos Vanrenterghem, Prof. Dr. Sabine Verschueren and initially Dr. Sam Van Rossom, later Dr. Danilo Catelli. The focus of this doctoral research project was the optimisation of knee loading in musculoskeletal rehabilitation through the objective quantification of structure-specific load parameters. Throughout his research project, Michiel had the opportunity to conduct multiple research visits to Cardiff Metropolitan University under supervision of Dr. Jasper Verheul.
Next to his research work, Michiel was also involved in clinical and educational activities. For four years, Michiel was a physiotherapist at the Sports Medical Advice Centre of the University Hospitals Leuven. Furthermore, he taught several theoretical guest lectures on biomechanics and was co-supervisor of multiple master theses and research internships. Apart from academic educational activities, he also engaged in educational activities for physiotherapists and the general audience, including LOKK peer reviews and the Dag van de Wetenschap.

List of publications

Articles in international peer reviewed academic journals

Hagen, M., Chebly, J., Dhaen, B., Fassian, N., Salvalaggio, M., Catelli, D.S., Verschueren, S., Vanrenterghem, J. (2024). Peak patellar tendon force progressions during heavy load single-leg squats on level ground and decline board. *Clinical Biomechanics*, *112* (106179). doi: 10.1016/j.clinbiomech.2024.106179

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Personal contribution

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