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Quantifying Muscle Forces and Joint Loading During Hip Exercises Performed With and Without an Elastic Resistance Band: Cumulative Master Thesis

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Abstract

Due to the rise in life expectancy and its resulting subsequent growth in the elderly population there has been a major increase in age-related pathologies. Many of these degenerative, non-communicable diseases, such as osteoarthritis (OA), are characterized by a rapid progression and have no known cure. It is therefore paramount that the efficacy of current rehabilitative measures be scientifically verified and research be conducted in order to develop new, innovative and effective methods.

To this end, the study was conducted to gain knowledge about muscle forces and joint loading during the performance of rehabilitation exercises used in a clinical setting to strengthen muscles important for joint stabilisation. The study specifically addresses exercises used in rehabilitation of OA of the hip using both the subject's own body weight and elastic resistance bands (ERB). In the study presented, hip joint kinematics, kinetics, muscle forces and hip joint contact forces (HJCF) during the individual exercises were determined via musculoskeletal (MSK) simulations using OpenSim and compared with those produced during walking. Finally, for validation purposes, HJCF were compared with those of a database based on in vivo measurements using instrumented hip implants.

In sum, the study found that ERB loaded exercises increased HJCF and targeted muscle forces. Furthermore, the study outcome shows that parameters such as ERB load and execution velocity of the exercises have minimal influence on variables such as peak muscle forces and HJCF. However, execution velocity does affect the total muscle force required for an exercise. Results of the comparison to walking show that hip exercises with or without an ERB resulted in lower muscle forces and lower peak HJCF during fast and slow hip extension and fast hip flexion exercises with no ERB loading.

This thesis contributes to OA rehabilitation research by providing systematic empirical evidence for clinically used hip muscle strengthening methods for rehabilitation purposes. It provides evidence that can be used either in further research or in a clinical setting for evidence-based rehabilitation training.

Zusammenfassung

Aufgrund des Anstiegs der Lebenserwartung und der daraus resultierenden Zunahme der älteren Bevölkerung haben altersbedingte Krankheiten stark zugenommen. Viele dieser degenerativen, nicht übertragbaren Krankheiten, wie z. B. Osteoarthritis (OA), zeichnen sich durch ein schnelles Fortschreiten aus und sind bislang nicht heilbar. Daher ist es von größter Bedeutung, dass die Wirksamkeit der derzeitigen Rehabilitationsmaßnahmen wissenschaftlich überprüft wird und Forschungen durchgeführt werden, um neue, innovative und wirksame Methoden zu entwickeln.

Zu diesem Zweck wurde die Studie realisiert, um Erkenntnisse über die Muskelkräfte und die Gelenkbelastung während der Durchführung von Rehabilitationsübungen zu gewinnen, die in einem klinischen Umfeld zur Stärkung der für die Gelenkstabilisierung wichtigen Muskeln eingesetzt werden. Die Studie befasst sich spezifisch mit Übungen, die bei der Rehabilitation von OA der Hüfte sowohl mit dem eigenen Körpergewicht als auch mit elastischen Widerstandsbändern durchgeführt werden. In der vorgestellten Studie wurden Hüftgelenkskinematik, Kinetik, Muskelkräfte und Hüftgelenkskontaktkräfte während der einzelnen Übungen durch muskuloskelettale Simulationen mit OpenSim ermittelt und mit denen beim Gehen verglichen. Zu Validierungszwecken wurden diese schließlich mit denen einer Datenbank verglichen, deren Ergebnisse durch In-vivo-Messungen mit instrumentierten Hüftimplantaten determiniert wurden.

Insgesamt ergab die Studie, dass ERB-belastete Übungen die Gelenkbelastung und die gezielten Muskelkräfte erhöhen. Darüber hinaus zeigen die Ergebnisse der Studie, dass Parameter wie die ERB-Belastung und die Ausführungsgeschwindigkeit der Übungen nur einen geringen Einfluss auf Variablen wie die Spitzenmuskelkräfte und die resultierende Gelenkbelastung haben. Die Ausführungsgeschwindigkeit beeinflusst jedoch die für eine Übung erforderliche Gesamtmuskelkraft. Die Ergebnisse des Vergleichs mit dem Gehen zeigen, dass Hüftübungen mit oder ohne ERB zu niedrigeren Muskelkräften und niedrigeren Gelenkbelastungsspitzenwerten während schneller und langsamer Hüftstreckung sowie schneller Hüftbeugung ohne ERB-Belastung führen.

Diese Arbeit leistet einen Beitrag zur OA-Rehabilitationsforschung, indem sie systematische empirische Beweise für klinisch verwendete Methoden zur Stärkung der Hüftmuskulatur zu Rehabilitationszwecken liefert. Sie liefert Beweise, die entweder in der weiteren Forschung oder in einem klinischen Umfeld für ein evidenzbasiertes Rehabilitationstraining verwendet werden können.

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1 Introduction

1.1 Life expectancy

The rapid advancement of technological development that has been ongoing in the last few decades is staggering (Roser, Ritchie, et al., 2013). Technological development as a concept is use of efficient technical means to facilitate intellectual work and increase productivity (Ruttan, 2000). As a continuous and ever accelerating process its effects are felt in every aspect of our day-to-day life, be that in form of faster and more efficient transport or the development of new and improved medication, such as a new vaccine. This advancement has in turn precipitated a large shift in the way we live (Ruttan, 2000). Arguably the most representative factor of the substantial impact that the increase in technological development has had on humanity as a whole is overall population health (Barclay et al., 2014; Crimmins, 2015; Omran, 2005). Through the advances in health care, public health and education, population health has seen a steady rise and nowhere is this clearer than when considering one factor: life expectancy (Crimmins, 2015; Roser, Ortiz-Ospina, et al., 2013).

While progress and development have undoubtedly led to a higher quality of life (QoL), perhaps the greatest impact is in terms of quantity of years lived rather than quality, especially considering that life expectancy as a variable is the key measure for assessing and quantifying population health (Roser, Ortiz-Ospina, et al., 2013). Given its importance, it is a testament to our the progress made in the last centuries that we have managed to increase our life expectancy from approximately 24 years in the 1800s to what it is today, over 80 years, in most first world countries (Bell & Miller, 2005; Crimmins, 2015; Roser, Ortiz-Ospina, et al., 2013). While the highest increases are in the highly developed first world countries, life expectancy is over 50 years even in the worst cases, i.e. third world countries. (Roser, Ortiz-Ospina, et al., 2013). For most of history, average life expectancy was relatively stable in relation to region. An average life expectancy of 35 years was a reality for most people just under 150 years ago, in countries that were highly developed for their time, while in extreme cases such as India and South Korea the average was even as low as 23 years (Crimmins, 2015; Roser, Ortiz-Ospina, et al., 2013).

While the doubling of our life expectancy can be attributed to several factors, one of the most obvious and probably one of the most drastic changes that have brought about this increase is the modernization and industrialization in the 1900s and the immense progress in the field of medicine and public health that proceeded it (Roser, Ortiz-Ospina, et al., 2013).

The introduction of publicly available medical resources has improved our ability to effectively combat acute diseases. This, together with the development of new vaccines and antibiotics, has been the reason why we have been able to effectively combat diseases such as smallpox and malaria, and reduce the infant mortality rate from around 50% to its current level (Roser, Ortiz-Ospina, et al., 2013). This marked the beginning of the steep and steady decline in morbidity, and subsequent mortality rates, due to communicable diseases, most notably infectious disease. (Armstrong et al., 1999; Roser & Ritchie, 2021).

1.2 The Epidemiologic Transition

This shift from a high mortality rate due to infectious diseases coupled with the ensuing rise in life expectancy is resulting in an age related demographic transition. The problem being twofold: (a) the increase in the number of the elderly population and (b) the progressive ageing of the older population itself. This trend of demographic aging can clearly be seen in recent decades, showing that the population group aged 70 or more is the fastest growing of any age group and projections of age related population changes show that the trend is likely to continue (Statistical Office of the European Communities, 2022). Statistical projections from Eurostat made in 2019 showed that the number of people aged 70 or older within the 27 EU countries is likely to increase by 56% in the next 30 years, from 63.7 million in 2019 to approximately 100 million by 2050 (see Figure 1).

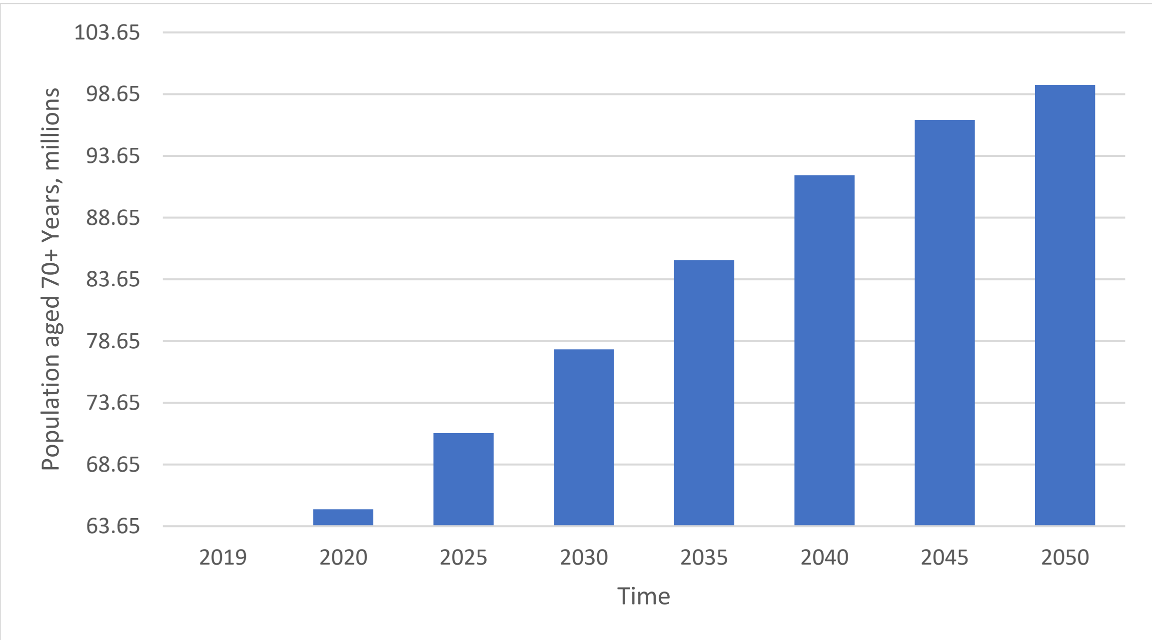


Figure 1: Demographic projection of the population aged 70 and over in the 27 EU countries from 2019 to 2050 according to the Statistical Office of the European Communities. Note: All data retrieved in July 2022.

This general upward trend, i.e. the increase in average life expectancy and the decrease in mortality rates, is an ongoing development that is likely to continue in the future (Crimmins, 2015). However, this age shift in the demographic of the population poses new problems that need to be addressed. With an increase in the elderly population a morbidity shift to more age related diseases, e.g. degenerative diseases, is inevitable (Armstrong et al., 1999; Brown, 2015; Roser, Ortiz-Ospina, et al., 2013; Roser & Ritchie, 2021). This conversion of the predominant morbidity type form communicable, e.g. infectious diseases, to more age related chronic, non-communicable diseases (NCD) is referred to as the “Epidemiologic Transition” from the "age of pestilence and famine" to the "age of degenerative and man-made diseases" (Omran, 2005).

1.3 Age-related Non-communicable diseases

The most common types of NCDs include cardiovascular and pulmonary diseases as well as other degenerative illnesses such as cancer, diabetes and various types of musculoskeletal (MSK) disorders. The main problem with NCDs is that while some can be non-infectious, acute illnesses that can be effectively treated, some are chronic-degenerative in nature and currently have no known cure. Thus, especially for chronic NCDs, the goal of health care shifts from acute treatment, as is customary for communicable diseases, to either prevention, such as reducing risk factors, or, in the case of disease onset, rehabilitation. Therefore, as further research in the fields of medicine and pharmacology focuses on the prevention of such diseases, the current ailments of the growing elderly population require effective, rehabilitative interventions to improve QoL and reduce the ever-increasing financial burden of chronic, NCDs, especially considering that such conditions result in particularly high rates of disability (Richards et al., 2016).

A showcase example of chronic, NCD include MSK disorders. According to (March et al., 2014) MSK disorders are only second to mental problems as the leading contributor to years lived in disability (YLD), a measure that indicates the extent to which a disease affects the QoL of the affected, before it subsides or, in a worst-case, leads to death. In this regard MSKs account for 21.3% all total YLD worldwide (March et al., 2014). In addition, many MSK disorders are characterized by having no cure and an accelerated progression (Gossec et al., 2005). Owing to this, these diseases need to be detected and treated as early as possible. Current clinical treatment consists of conservative, non-invasive therapies or, in most cases despite conservative measures, with invasive surgical procedures that are not only very costly but also pose a significant risk to patients. This creates a high motivation to control the progression and development of such disease patterns with the improved conservative, non-pharmacological treatments as effectively and early as possible (Nho et al., 2013; W. Zhang et al., 2008). Furthermore, due to the characteristic pain and mobility restriction caused in joints, bones and muscles by MSK disorders, these types of clinical conditions not only result in a significant restriction of the ability to work, which leads to premature retirement from working life, but also poses a considerable hurdle to the effective participation of those affected in social life which in turn leads to a substantial reduction in the QoL (Roux et al., 2005). As a consequence of the above-mentioned sequelae of MSK diseases, there is a high motivation to curb the progression and the associated symptoms effectively by rehabilitative measures.

1.4 Osteo arthritis

One such MSK condition is chronic osteoarticular disease, or osteo arthritis (OA). OA, the most prevalent form of arthritis, is a degenerative joint disease that, unlike other forms of arthritis, is not primarily inflammatory (Cross et al., 2014; March et al., 2014). The disease is characterized by a progressively, degenerative change in cartilage and bone structure, tending to occur mainly in the hands, neck, lower back, knees and hips, that can eventually lead to joint deformity (Glyn-Jones et al., 2015). It results primarily from many years of over-use, increased joint loads and/or a focal distribution of joint loading. Felson (2013) identifies a number of primary factors that can lead to the development of increased local joint forces. He further proposes that these are the major causes for the development of OA: (1) A focusing of the otherwise normal total joint load caused by congenital or acquired irregular or abnormal anatomy, (2) Injury caused by athletic or other similar activities that results in an acute or chronic overload, (3) Increased total stress brought on by chronic obesity, and (4) A combined pathology of abnormal anatomically induced form and the occurrence of increased stresses.

Due to its detrimental effects on the protective cartilage and underlying bone of the afflicted joint it has a significant impact on mobility and function, which in turn has a considerable negative influence on the QoL of those affected and can, in later stages of the disease, lead to impairment of many daily activities (Neogi, 2013). Due to its progressive nature, if left untreated, OA can rapidly lead to disability. According to March et al. (2014), AO is a significant contributor to YLD, showing that OA was the cause of 10.5 % of YLD in 2010 due to MSK disabilities, which in turn in is a large financial stress on health care systems. This further demonstrates that the resulting burden of various forms of OA account for a significant proportion of the MSK-related disabilities (Cross et al., 2014).

In regards to locality, the two most prevalent forms of OA are knee and hip OA (Glyn-Jones et al., 2015). These two joints in particular are often affected due to them being two of the highest load baring joints in the human body and thus prone to afflictions that are the result of overuse.

A further problem seems to be the triggering of so-called pathomechanical movements or gait strategies due to OA progression. These alterations in kinematics and kinetics are evasive or compensatory in nature with the goal of relieving joint load of the affected joint as much as possible, which in turn avoids pain and compensates for muscle weakness (Meyer et al., 2018). Unfortunately, these pathomechanical movements seem to be one of the problems with regards to further OA progression. Felson (2013) even argues that from the time of the appearance of such pathomechanical movement forms, that the pathomechanics outweighs all other factors in causing the disease progression. This occurs due to changes in cartilage and bone structure of the effected joint, resulting in a more focal stress distribution. In addition, compensatory movement strategies can also lead to significantly higher loads on other joints not affected by OA in order to counterbalance the load-relief of the OA affected joint. As a result, other joints are subjected to higher loads which in turn could precipitate a more rapid degenerate, exacerbating the problem and increasing the likelihood of further pathologies developing due to overuse or overload.

Due to the aforementioned tendency of MSK disorders to an accelerated progression with age, hip and knee OA are also the most common reason for surgical intervention, such as a total endoprosthesis (TEP) hip and knee replacement, after just two years (Gossec et al., 2005; Y. Zhang & Jordan, 2010). This, in turn, serves to highlight the importance of conservative measures to counteract and slow down the progression of the disease.

1.5 Treatment of OA

Regarding OA treatment, the treatment options are categorized twofold: (1) Surgical treatments and (2) non-surgical treatments.

In the surgical treatment of osteoarthritis of the hip, a distinction can be made between two types of surgical strategies: (1) joint-preserving surgery and (2) joint replacement with a hip prosthesis (de l’Escalopier et al., 2016). The most important joint-preserving surgical procedure is the so-called osteotomy. In this procedure, an attempt is made to preserve the hip joint by rearranging the joint axis. To do this, the hip socket and the thigh bone are rearranged so that the load-bearing surface in the hip joint is changed in such a way that the main load rests on parts of the hip joint that are still healthy. In hip joint replacement, the diseased hip joint is partially or completely replaced by an artificial hip joint, i.e. a hip TEP.

Endoprostheses is a replacement of the hip joint whereby the components of the natural hip joint are replaced by artificial materials. Endoprostheses can also be used as partial replacements and in different joints. The hip TEP, however, is a total replacement of the joint head of the hip with an artificial implant. The hip TEP is one of the most common types of invasive treatments for hip OA and is generally complication-free, but as with all operations, infections, nerve injuries, post-operative bleeding or blood clots can occasionally occur (Kristensen et al., 2014; Kunutsor et al., 2016; White & Henderson, 2002). While modern prostheses have a very long lifespan, with 87.9% of artificial hip joints still functioning properly after 15 years, premature loosening of the implant can sometimes occur, and the earlier the surgery is performed, the more likely it is that the prosthesis will need to be replaced at some point (Evans et al., 2019). As a rule, the indication for hip TEP should be given if the patient reports high subjective suffering pressure with regard to hip-related complaints, namely pain, functional limitations, restrictions on activities of daily living and health-related quality of life, despite previous conservative therapy (German Society for Orthopaedics and Trauma Surgery, 2021).

Until these indications are given or if surgery is not possible or recommended, e.g. previous and active infections, acute or chronic concomitant diseases or BMI ≥ 40 kg/m², non-surgical OA treatment options, i.e. conservative treatment methods, are necessary (Kristensen et al., 2014; Kunutsor et al., 2016; Lenguerrand et al., 2018; Pugely et al., 2015; Radtke et al., 2016). Conservative treatment options are further broken down into pharmaceutical and non-pharmaceutical treatments. In mild to moderate cases of OA medication such as topical or oral non-steroidal anti-inflammatory drugs (NSAIDs) are often used in conjunction with pain medication such as paracetamol (Conaghan et al., 2008; Nelson et al., 2014). In severe cases steroidal treatments are used to induce short-term pain relief but regardless of the anatomical site, the pharmacological treatments are generally uniform and the rehabilitation is tailored to the individual depending on the effected joint (Rannou & Poiraudeau, 2010).

Aside from weight loss in cases in which it is deemed necessary, non-pharmaceutical treatments include self-directed exercise as well as physiotherapy. Physical therapy is typically, in contrast to self-directed exercise, a form of planned and structured physical exercise, in which modality and control measures are usually provided and directed by physiotherapists (Rannou & Poiraudeau, 2010). In fact, according to the OA Research Society International (OARSI) recommendations (2014), rehabilitation, including various forms of exercise therapy, is considered a core treatment for OA and is recommended for all patients.

In physiotherapy, there are many types of exercise modalities that are used to treat OA. These modalities include both aerobic activities and muscle strengthening exercises, both of which have been shown to significantly improve mobility and provide significant and lasting relief from the pain of affected joints (Ettinger et al., 1997; Nelson et al., 2014). Apart from the age of the affected patient, the choice of modalities used during therapy is largely dependent on the affected area of the OA as well as the grade and severity of the disease progression, whereby the choice of therapy should also be adapted to possible accompanying comorbidities (Nelson et al., 2014).

Furthermore, as mentioned in the chapter above, compensatory movement strategies also seem to occur in patients with hip OA. According to Meyer et al. (2018), these compensatory movements are often also a result of muscle weakness in the surrounding hip muscles. The findings of these occurring muscle weaknesses is also further confirmed by Loureiro et al. (2013) in a systematic review. They show that patients with hip OA have significantly less muscle strength, muscle size and muscle quality than in the leg not affected by OA, or indeed in contrast to a healthy control group. This in turn reinforces the approach that strengthening the weakened musculature of the affected leg through targeted muscle training can counteract such frequently occurring imbalances in order to both limit the progression and further improve the QoL of patients (Zhang et al., 2008; Nho et al., 2013). It could also be argued that in view of the average age of the patients suffering from OA, muscular rehabilitation is of particular importance, as a common comorbidity at this age is suffering from muscle atrophy, so-called sarcopenia. Research has shown that not only muscle strength but also muscle mass declines considerably with age (Brooks & Faulkner, 1994). According to one study, 80-year-olds have a 40% decrease in muscle mass compared to people in their twenties (Kalyani et al., 2014). While the effects of sarcopenia on OA onset and development are as of yet unclear, it is theorized that with natural onset of muscle atrophy in old age, OA could become a greater risk factor or, in the case of OA that has already occurred, could lead to a significant progression (Papalia et al., 2014). This, in turn, would lead to less movement of the patients and increase the likelihood of other joints becoming diseased.

Strengthening the hip stabilizer muscles improves the stability of the joint and reduces joint contact forces (Retchford et al., 2013; Meyer et al., 2018). In other words, strengthening the pelvi-trochanterian muscles surrounding the hip could increase stability due to a more balanced muscle force distribution which reduces femoral head translation and therefore decreases joint contact forces (Nguyen et al., 2016).This is especially critical because the presence of increased HJCF is one of the main contributing mechanical causes of hip OA and its progression (Recnik et al., 2009; Felson, 2013). Therefore, the knowledge, understanding, and subsequent control of these forces are essential for building a progressive rehabilitation program.

The muscle stimulus needed to strengthen muscles can be achieved with different exercise modalities (Hofmann et al., 2016; Iversen et al., 2018). In strength training, there is the possibility of exercising with one's own body weight or using external loads to provide a significant stimulus to the concerned musculature. While the use of classic strength training devices or free weights is common practice, the problem remains that such equipment is only available in specialized environments such as gyms and/or the equipment is very expensive. However, there are cost-effective training devices that provide an alternative. One of them is the use of an elastic resistance band (ERB) to achieve the necessary training load. ERBs are easy to use, easy to store and come in a variety of strengths. They can be used in a multitude of ways to exercise any muscle group needed in OA rehabilitation. Because of these many benefits and due to the body of literature that advocate moderate intensity muscle strengthening exercises, progressive resistance training using elastic resistance bands is commonly used by physical therapists in exercise therapy as an effective means to treat hip OA (Hofmann et al., 2016; Iversen et al., 2018).

When designing an exercise plan that is rehabilitative in nature, various exercise management parameters must also be considered. One of these being that both amount and extent of work performed by the patient during therapy must be adapted to the patient's current condition. This includes controlling the resistance or load of the exercises, e.g. the weight of the exercises or the strength of the ERBs, in order to create an effective stimulus for muscle growth but not exacerbate condition symptoms (Feigenbaum & Pollock, 1999; Häkkinen, 2004; Helms et al., 2019). Furthermore, the frequency of the rehabilitation sessions must also be taken into consideration. The sessions must be set close to each other to ensure progressive muscle development, but not so close that it interferes with the recovery process of the last session (Helms et al., 2019). The duration of the sessions and the individual progression of the patient should also be continuously adjusted to the current level of conditioning. It is also necessary to choose the type of exercise and its implementation appropriately in order not to compound any symptoms arising from OA or in some cases any other comorbidities (Feigenbaum & Pollock, 1999; Häkkinen, 2004). In this regard, the environment in which the rehabilitation will take place should be considered. Whether the sessions are to be supervised or unsupervised, in a gym or in the patient's home, will largely determine the choice of exercise. However, for rehabilitation recommendations, personal preferences should be strongly considered during programme design in order to achieve the highest possible patient adherence, similar to general and preventive physical activity recommendations (Aboagye, 2017). This is particularly important as the patient's attitude towards the recommended rehabilitation exercises and the likelihood of performing them in an unsupervised environment are of utmost importance for the success of the rehabilitation (Teo et al., 2022).

In this context, there also remains uncertainty about the appropriate training intensity for strength training in OA. While greater training effects have been demonstrated in individuals who engage in high-intensity strength training, there is concern that high joint loading to achieve these intensities may increase pain and joint stress in joints affected by OA (Regnaux et al., 2015). This in turn raises the question of whether higher intensity exercise routines might have more detrimental long-term effects compared to performing the same exercises at a lower intensity.

Although physical therapy, including muscle-strengthening exercises, is a common recommendation for the clinical treatment of knee, hand, and hip OA, the state of the research, and thus the evidence, varies widely depending on the affected joint (Nguyen et al., 2016). Most of the evidence for OA recommendations is based on studies that primarily examined patients with knee OA. While these studies have shown that exercise therapy and specific exercises to strengthen lower limb muscles improve mobility and function while leading to pain reduction, there is a lack of research on OA in the hip related to rehabilitative exercises (Ettinger et al., 1997; Häkkinen, 2004; Nguyen et al., 2016; Regnaux et al., 2015). This has led to the current rehabilitative exercise recommendations to treating therapists as well as patients suffering from hip OA being based on the results of studies related to knee OA. This means that in order to ensure the efficacy and safety of exercise therapy and strength training in rehabilitation, it is of significant importance that studies address the effects of the exercises used in practice for the rehabilitation of hip OA specifically, especially in a biomechanical context, to ensure that these recommendations are effective in combating the disease being treated.

Considering that interventions utilizing lower-limb hypertrophic strength training through resistance-based exercises have been shown to have been an integral part of conservative hip OA rehabilitation, as they have been shown to result in an increase in joint stability, there seems to be surprisingly little research focusing on expanding on, or validating current clinically used methods and exercises (Meyer et al., 2018; Nguyen et al., 2016; Retchford et al., 2013; W. Zhang et al., 2008). Taking into account that both the patient and the treating physiotherapist depend on certain exercise recommendations, it is important that these recommendations are based on scientific evidence. However, this does not seem to be the case. In most cases, the choice of therapeutic exercises, as well as the associated control measures, is based on expert opinion (Conaghan et al., 2008; W. Zhang et al., 2008).

1.6 Joint contact forces

In this context, joint contact forces (JCF) seem to play an important and integral role. In relation to causes of OA and its progression, research has shown that the presence of increased contact forces in the hip joint (HJCF) is one of the most significant mechanical causes (Felson, 2013; Recnik et al., 2009). For this reason, the detection, management, and successful manipulation of these stresses are paramount in order to successfully design a scientifically based, long-term rehabilitation program in order to curb the progression of pre-existing pathologies.

In general, JCFs are defined as the forces generated in a joint by articulating surfaces in response to loads acting on the said joint. It results from the need to balance the moment arms of the body weight and the surrounding loads to allow the maintenance of desired body positioning. The size and magnitude of the aforementioned JCFs result from and are dependent on the influence of three main factors: (1) external forces acting on the human body, e.g. due to movement or acceleration of a particular body part, (2) the total body weight of the subject and (3) the active and/or stabilising musculature of the respective joint at that given moment (Becker et al., 2020).

This would then mean that the higher the muscular forces of the surrounding muscles as a result of the training, the higher the HJCF. For example, higher anterior traction muscle forces would lead to a higher HJCF in the posterior direction (see equation 1 for HJCF below). However, an increase in joint stability could lead to a more balanced distribution of muscle forces (e.g. between anterior and posterior traction muscle forces) and thus reduce HJCF. In other words, increasing joint stability could reduce femoral head translation and thus HJCF.

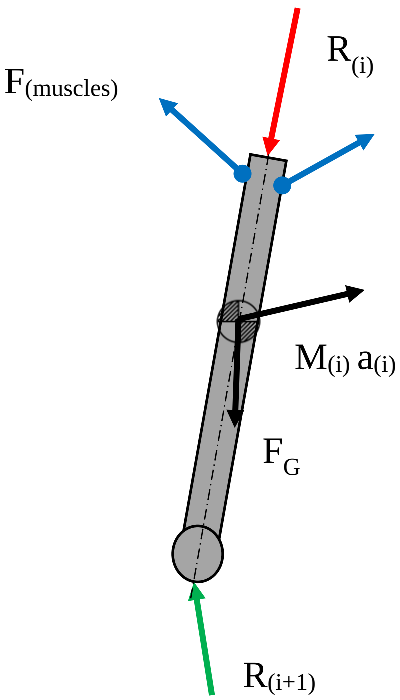


Figure 2: Representation of a thigh segment with acting forces for calculating the proximal joint force, i.e. hip joint force. Whereby $R_{(i)}$ = Proximal joint force, $M_{(i)}$ = mass of segment, $a_{(i)}$ = acceleration of segment, $F_{(external)}$ = External forces, $F_{(muscles)}$ = Sum of all muscle forces which act on the segment, $R_{(i+1)}$=Joint force in the distal segment, in this case of the knee

$$\vec{R}_{Hip} = [M]_{femur} \times \vec{a}_{femur} - (\vec{R}_{knee} + \vec{F}_{gravity} + \sum \vec{F}_{muscles_{femur}}) \tag{1}$$

$\vec{R}_{Hip}$ ... proximal joint force

$[M]_{femur}$ ... mass of the femur

$\vec{a}_{femur}$ ... acceleration of the femur

$\vec{R}_{knee}$ ... Joint force in the distal segment i.e. knee

$\vec{F}_{gravity}$ ... External forces

$\vec{F}_{muscles_{femur}}$ ... Sum of all muscle forces acting on the femur

1.7 Motivation and problem statement

In view of the above-mentioned knowledge gap, the aim of publication P1 was to biomechanically analyse practical hip stabilizing muscle exercises by further analysing the internal HJCF during the exercises to hopefully substantiate the movement recommendations in terms of exercise selection and execution. Furthermore, in order to put the data obtained from the analysis in a generally relevant relation, the HJCF and other biomechanically relevant variables measured during the exercises were compared with those found during an ordinary gait cycle by the same participant. To validate the HJCF obtained from the MSK simulations, the results of the simulations were additionally compared to the HJCF obtained from in vivo measurements using instrumental implants.

In order to make the exercise selection of the analysed movements as practical and patient-relevant as possible, the loading modality chosen was the use of body weight and the use of ERBs. This choice guarantees that the conclusions drawn will not only be relevant for expert use in outpatient or inpatient rehabilitation, but also relevant for patient recommendations to perform independent rehabilitation.

2 Methodology

While movements such as walking and stair climbing have been studied both in vivo, using instrumented implants, and through MSK simulations, the recommended rehabilitation exercises have not been analysed as extensively (Stansfield, et al., 2003). While some movements and their resulting HJCF have been characterised in vivo, a comparison with those resulting from MSK simulations would be of interest to serve as a reference for further research. (Schwachmeyer, et al., 2013). Study P1 focuses on analysing the HJCF of a range of clinically recommended rehabilitation exercises for the purpose of obtaining important clinically relevant data on these exercises.

Sixteen healthy volunteers had their movements recorded using force plates, EMG (Electromyography) sensors and a camera-based movement analysis system (VICON). For this purpose, 21 three-dimensional retro-reflective surface markers & 5 trilateral marker clusters as well as 6 wireless EMG electrodes were placed on predefined anatomical landmarks to record the movement and muscle activity of the lower extremities. All examinations were carried out in the biomechanical laboratory of the Institute of Sport Science in Vienna. After a gait analysis, which served as the basis for comparison, all subjects were asked to perform a series of simple, non-intensive exercises. The data was then used to create MSK simulations of the movements using open source biomechanical modelling software (OpenSim). These simulations in turn allowed the HJCF to be determined and compared for each of the exercises. For perspective, the resulting HJCF of the exercises were compared to those of an ordinary gait cycle.

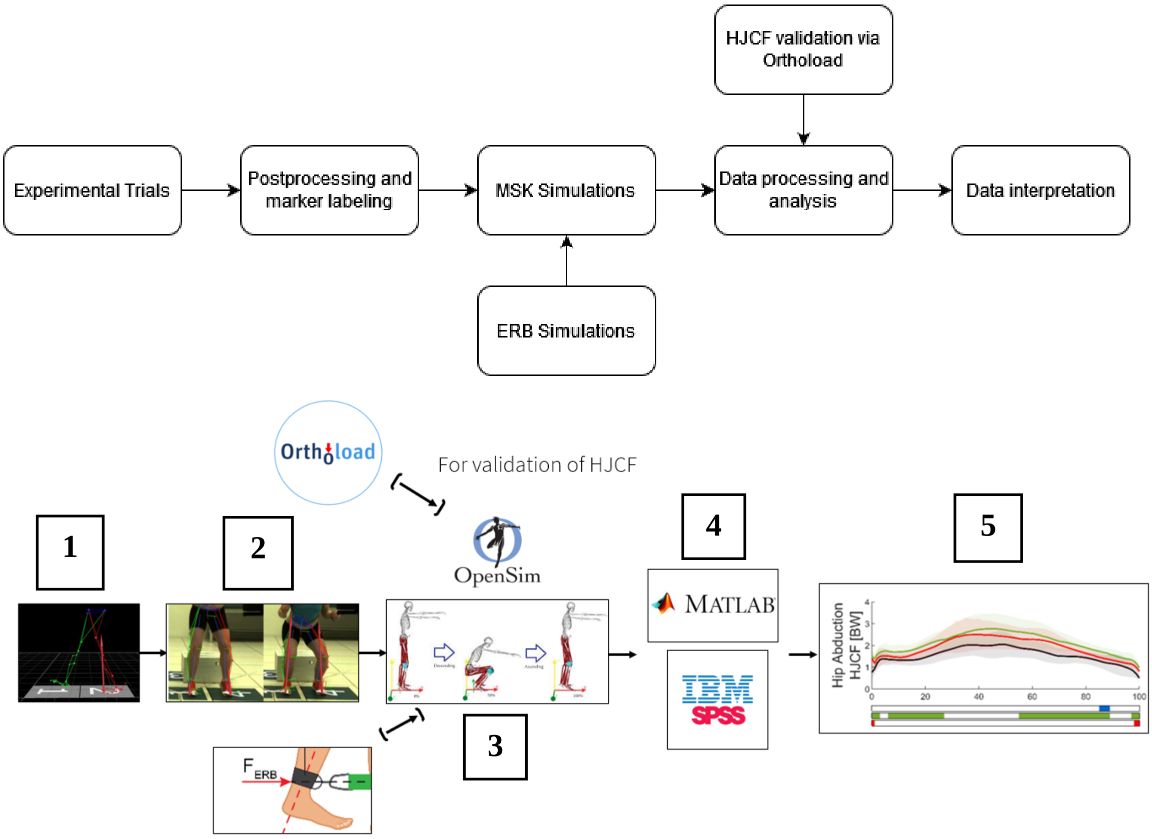


Figure 3: Schematic representation of the experimental procedure and the various steps of the post-experimental data processing. (1) Experimental trails were conducted with each subject, (2) marker labelling in nexus, (3) hip joint kinematics, kinetics as well as hip contact forces were calculated based on MSK simulations (OpenSim), (4) and (5) the time-normalised waveforms of the different execution modalities were compared with each other (using statistical parametric mapping analysis) as well as with those found during walking. For validation purposes, a visual comparison of our waveforms with those of a public database (Orthoload) using instrumental hip implants was also performed.

2.1 Participants

The participants were recruited through direct contact or postings. The study participants were therefore mainly students or employees of the Institute of Sport Science, the exclusion criteria being any pre-existing pathologies as well as being an age below 18 or above 45 years. No individuals from protected groups were included in the study.

The sample consisted of 11 male and 5 female healthy volunteers. Their average ± SD (Standard deviation) age, weight, height, and body mass index were 27 ± 4 years, 70.7 ± 12.5 kg, 1.75 ± 0.10 m, and 22.9 ± 2.8 kg m^{-2}, respectively (see Table 1 and Table 2). All subjects were free of injury, pain and neuronal defects at the time of data collection. In addition, any lower limb injuries must have been at least one year in the past at the time of data collection.

Table 1: Anthropometric data including age, mass, height, inter-anterior superior iliac spine (ASIS) distance, sex and BMI of each of the 16 participants.

ID	Age	Mass [kg]	Height [cm]	Inter-Asis [cm]	Sex	BMI [kg m^{-2}]
p01	26	51	162	21.4	Female	19.4
p02	23	67.5	180	24.5	Female	20.8
p03	35	55.6	172	22	Female	18.8
p04	33	66	165	23.5	Male	24.2
p05	29	75.8	175	23.1	Male	24.8
p06	26	87	195	27.9	Male	22.9
p07	27	88.7	178	27.4	Female	28.0
p08	29	89.5	192	28	Male	24.3
p09	23	85	182	25	Male	25.7
p10	22	69.9	180	24	Male	21.6
p11	35	61	159	25	Female	24.1
p12	25	55.3	175	24	Male	18.1
p13	27	81	177	26	Male	25.9
p14	24	78.4	178	23.6	Male	24.7
p15	24	66	174	23	Male	21.8
p16	28	70	170	25	Male	24.2

Table 2: Anthropometric data showing minimum and maximum values as well as mean values and standard deviations.

	N	Min.	Max.	Mean	SD
Age [J]	16	22	35	27	4.1
Mass [kg]	16	51.0	89.5	70.7	12.4
Height [cm]	16	159.0	195.0	175.6	9.5
BMI [kg/m2]	16	18.1	28.0	22.9	2.8
Inter-asis [cm]	16	21.4	28.0	24.5	2.0

2.2 Ethics

In terms of research ethics, all the necessary aspects of participant safety and data protection were considered and taken into account. Participants were explicitly informed of the possibility to withdraw from the study at any time before the start of the study. Participants were also informed that they could contact the investigator regarding their withdrawal at any time by telephone, email or in person without giving any reason. For the purposes of the study, personal data such as name, height, weight, sex and inter-asis distance were collected prior to the measurements. Furthermore, the participants were not beholden in any way to the study coordinator or the study director. There is no relationship of dependence between the participants and the investigator, even if the participants are students of the Institute of Sport Science, as participation in this study does not lead to any study-related benefits.

To ensure data protection and anonymity, the personal data of all participating participants were stored in a sealed paper form. A digital form of the personal data, anonymised with numerical codes, was stored separately from the measurement data, which was also anonymised, on different data carriers (computer in the biomechanics laboratory of the University of Vienna). The corresponding numerical coding was stored on a third data carrier (external, password-protected USB stick), to which only the project leaders had access. In addition, all data carriers were stored in the biomechanics laboratory of the University of Vienna. The lab is locked at all times and only accessible with an electronic ID card. All laptops or computers on which the subjects' personal data were stored used a personal, password-protected log-in. If personal data are stored on external data carriers or documented in paper form, the devices or documents are kept in a locked room to which only employees of the Department of Biomechanics, Movement Science and Sports Informatics at the University of Vienna have access. The participants had the right to personally ask the project leader to inspect their data at any time. Participants had the option to withdraw their data at any time by telephone, email or in person without giving a reason.

During data collection, participants were not exposed to any risk of injury and if pain or discomfort would occur, it was possible to discontinue participation in the study at any time. Since the risk factor of the experimental procedure was almost non-existent in this study and the potential knowledge gained is relatively high, it stands to reason that the ratio of potential risks to expected scientific and societal benefits is very much in favour of the expected scientific and societal benefits as new findings can later provide important information for sports science and therapeutic issues.

An interruption of the study was foreseen in case more than half of the subjects would drop out during the study. In this case, the reason for the dropout would be recorded and taken into account for further studies. In addition, the study would have been interrupted if the conduct of the test had resulted in a health risk for the subjects. In the case of a failure or malfunction of measuring equipment that could not be remedied within a reasonable period of time, or which required a high cost contribution, the study would also have been discontinued.

The research ethics and methods of the study were approved by the Ethics Committee of the University of Vienna (00579), and all participants were informed regarding the purpose of the study and gave their written consent before participation

2.3 Data collection

In the first step of data collection, the participants were informed about the purpose of the study and the procedure. They were also given the opportunity to ask questions or raise concerns about the study and its conduct, and to discuss them if necessary. They were also informed about the risks and data protection measures described in the ethics section of the paper and confirmed their consent with their signature. Then name and anthropometrical data such as height, weight and sex were noted and the inter-asis distance was measured and recorded for later use.

2.4 Exercises

The choice of exercises measured had to be adapted to certain criteria. For this purpose, the exercises had to be chosen in such a way that (1) they target the stabilising muscles of the hip, (2) they can be performed in such a way that both the left and right leg are allocated to one force plate to ensure the absorption of the reaction forces, (3) they can be designed as similar as possible to ensure a meaningful comparison, (4) they can be performed with both ERB and body weight without altering the movement, (5) they do not pose a risk of injury to the participant, and (6) the exercises used in the study are those used in the practice of hip OA rehabilitation.

Thus, in relation to point (1), the exercises had to target the following hip stabilising muscles: (a) hip abductors including gluteus medius, gluteus minimus, tensore facie latae, and piriformis (Valente et al., 2013; Meyer et al., 2018); (b) hip flexors including rectus femoris, iliacus, psoas, iliocapsularis, and sartorius (Zhang et al., 2008); and (c) hip extensors including gluteus maximus, biceps femoris, semitendinosus, and semimembranosus (Loureiro et al., 2013).

Ultimately, it was decided to use five difference exercises:

(1) bodyweight squat, as it has been shown to significantly increase the strength and size of the lower body muscles and also develop core strength (Lorenzetti et al., 2018). In addition, the squat is a rehabilitation exercise often used in practice and is very often part of a lower extremity muscle building programme as it is such an effective multi-joint exercise that uses a variety of leg muscles (Escamilla, 2001).

(2) a standing single-leg abduction to target the aforementioned hip abductor muscles,

(3) a standing single-leg hip extension to target the aforementioned hip extension muscles, and (4) a standing single-leg hip flexion to target the aforementioned hip flexor muscles. Exercises

(2)-(5) were also specifically chosen because of the similarity in their execution. They are all one-legged exercises performed in a standing position in which the non-performing leg could always be in contact with the recording force plate. Furthermore, with all the above-mentioned exercises, the application of an ERB to increase resistance is possible and can also be performed by OA affected patients.

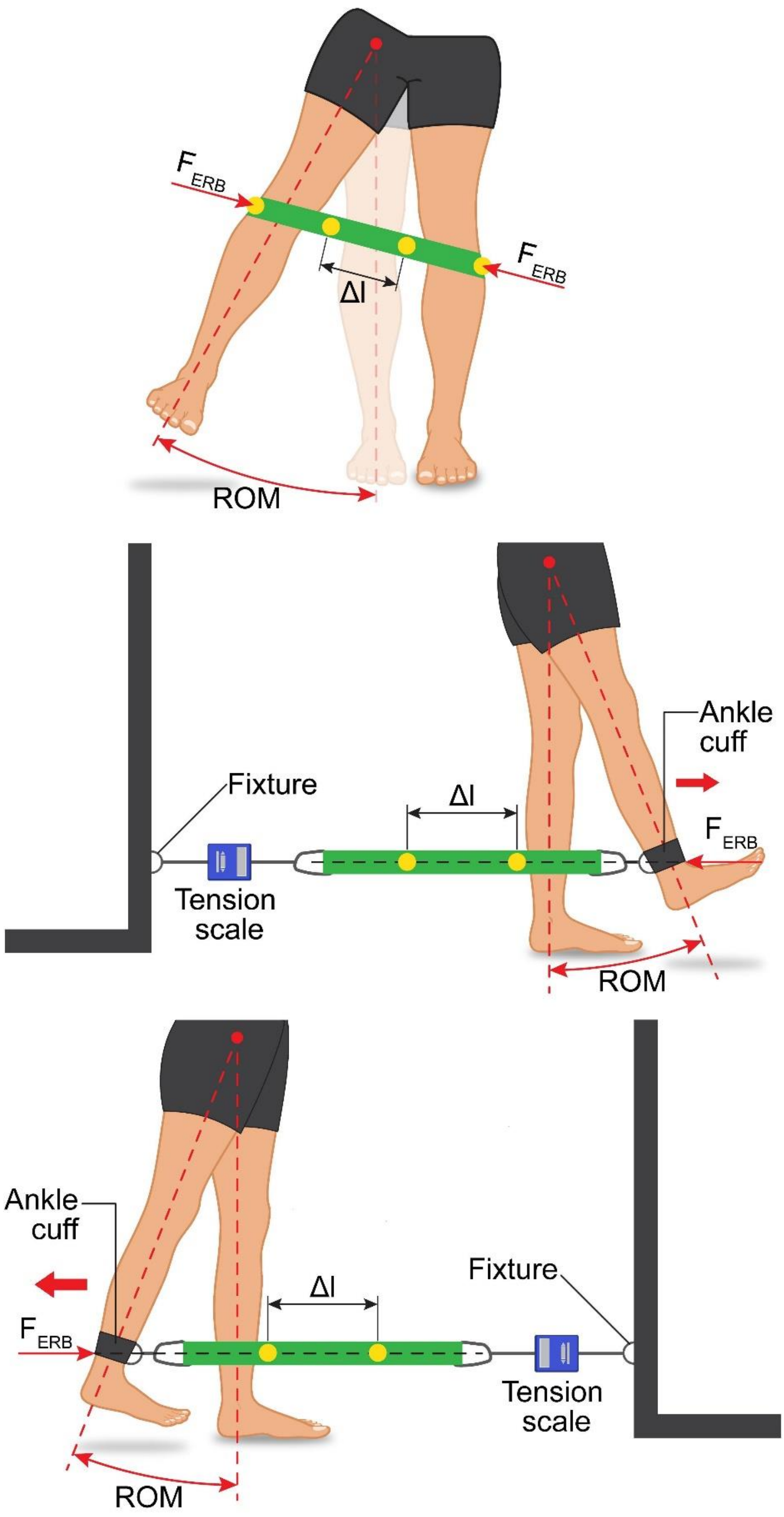


Figure 4: All three variants of the experimental set-ups using an ERB showing exercise execution and fastening method for the ERBs. (1) standing single-leg abduction, (2) standing single-leg hip extension and (3) standing single-leg hip flexion. Marker displacement (Δl)

was using during MSK simulations in order to calculate the force production of the ERB (F_{ERB})

2.5 Elastic resistance Bands

As mentioned above, it was decided to use ERBs to apply the load to the two exercise conditions that were to be exercised against resistance. The ERBs used in the publication P1 are from the brand TheraBand (Thera-Band, OH, USA), as they seem to be the most common brand of ERBs. ERBs are generally resistance bands made of latex rubber. Due to their elasticity, pulling on the bands creates a resistance that can be used during exercise to provide a muscular stimulus. They are normally implemented by users to rehabilitate injuries as well as enhance athletic performance. The advantages of ERBs are that, unlike other resistance modalities, they are very easy to use and very affordable.

In view of the choice of resistances offered by the TheraBand company, it was decided that a red and a green TheraBand would be used in the study. As can be seen in Table 3, the red ERB from TheraBand is the softer of the two and the green one is the stiffer one, which, according to the company's description, should cause a 25% higher resistance at 100% elongation. The 25% increase was considered to be a large enough difference in load to produce different results in the experimental trials.

Table 3: The difference in force production between the different variants of ERB made by TheraBand

Band Colour	**Increase from preceding colour at 100% Elongation [%]**	**Resistance [kg]**		**Resistance increase from 100%-200% [%]**
		at 100% Elongation	**at 200% Elongation**	
Tan	-	1.09	1.54	41.67
Yellow	25	1.36	1.95	43.33
Red	**25**	**1.68**	**2.49**	**48.65**
Green	**25**	**2.09**	**3.04**	**45.65**
Blue	25	2.63	3.90	48.28
Black	25	3.31	4.63	39.73
Silver	40	4.63	6.94	50.00
Gold	40	6.44	9.66	50.00

Note: All values taken from the TheraBand website (TheraBand, 2022)

The company describes the red ERBs as "medium heavy" and its intended use as particularly suitable for women, adolescents and trained seniors, whereas the green one is described as "heavy" and would be particularly suitable for less trained men, trained women and adolescents. These descriptions were felt to be appropriate to provide a necessary muscular stimulus for the intended target group.

Given the high number of exercise cycles the ERBs had to withstand, and their elastic nature, it was felt appropriate to ensure the consistency of the ERBs tension characteristics throughout the experiment by testing them for different force development before and after each subject. To verify the assumption of a linear relationship between force and elongation, both ERBs used were evaluated in an experimental test. Hence, to prove that the ratio of strain to force was the same in all tests, a number of different weights were attached to the ERB and the strain was measured using the Vicon system. For this purpose, the stiffer ERB was loaded with 0, 0.5, 1.0, 2.5, 5 and 7.5 kg, and the softer ERB with 0.0, 0.5, 1.0, 2.5 and 5 kg. The displacement of the attached reflective markers was measured and then used to fit a line to the experimental force and strain data.

The length measurements for the static loading experiment, where displacement was evaluated, were measured twice over two different predefined marked distances on the respective ERBS, once with a calliper and again using fixed reflective surface markers whose displacement was measured with a Vicon system.

During the mapping of our ERBs force-extension curve it became apparent that elastic hysteresis was hardly noticeable within the curve the data points produces. The curve looked to be very close to linear and using a linear approximation to plot our force-extension curve would made the subsequent computations easier. In light of this, a comparison was conducted in order to evaluate whether or not using this linear curve would yield significantly different results. Therefore, four different curve-types were fitted to the experimental data: a) a force-elongation curves based on the assumption of a linear relationship, b) a force-elongation curve based on a 2nd degree polynomial curve, c) curve based on a 3rd degree polynomial curve, and d) curve based on a 4th degree polynomial curve. While the results of the comparison showed that the 2nd degree polynomial curve would have been a better fit, a statistical analysis was conducted in which the findings showed that there was no significant difference in results while using a linear approximation.

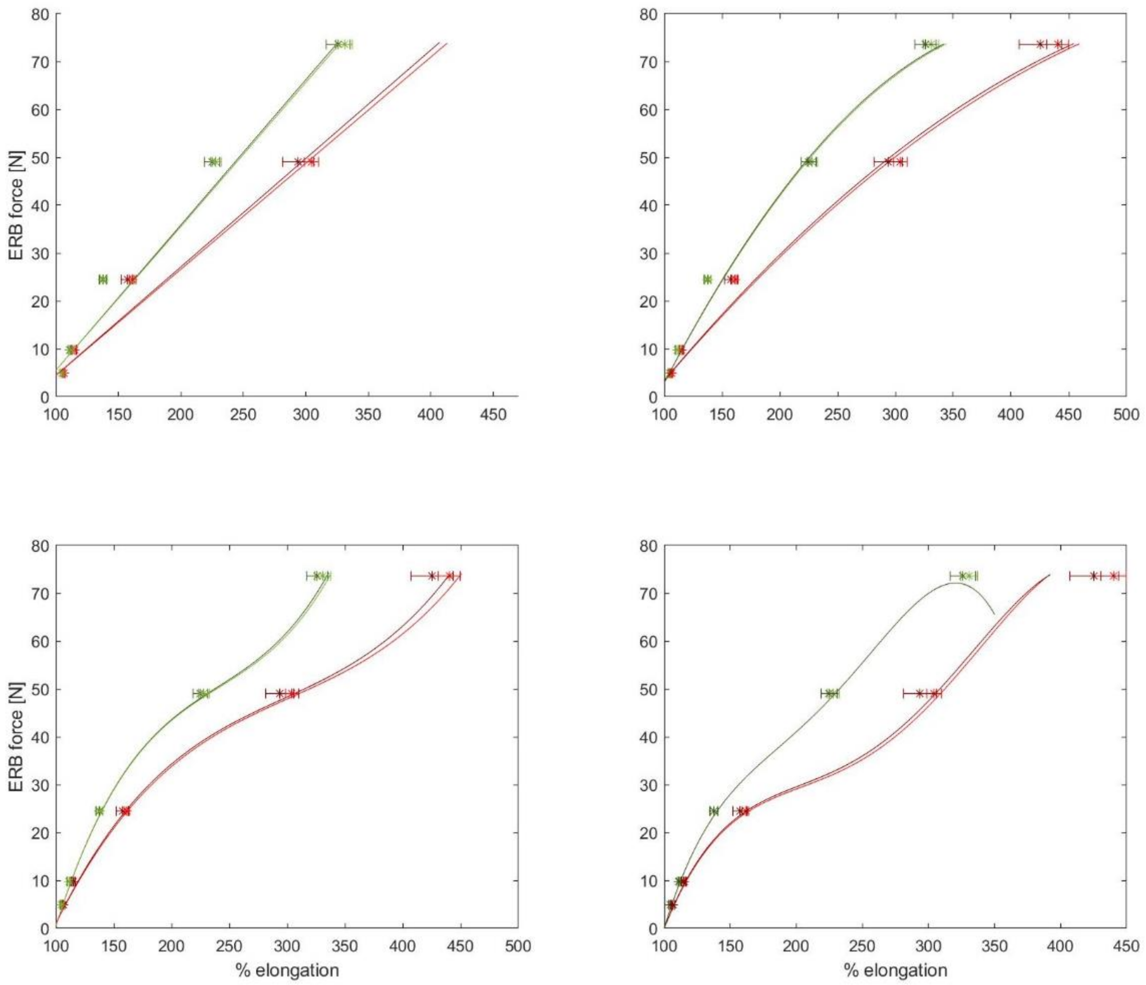


Figure 5: Mean force-elongation curves based on the assumption of a linear relationship (left top plot) and non-linear relationship. Right top plot: 2nd degree polynomial curve, Button left plot: 3rd degree polynomial curve, Button right plot: 4th degree polynomial curve.

2.6 Experimental setup and procedure

During parts of the experimental procedure in which the ERBs were used they needed to be fixed to the participant in order to counteract the point of force application of the ERB resistance from shifting during the trials. Thus, during the flexion and extension trial, in which the movement of the ERB was considered likely, an ankle cuff was fitted to the participant. The ERB used in the specific trial was then attached to the ankle cuff as well as to a static fixture. Furthermore, the fixture was aligned with the moving leg, in order prevent the generation of forces that do not point in the direction of movement as much possible.

In order to keep the point of force application of the ERB resistance as similar to those seen in a clinical use as possible, in the case of extension and flexion exercises using ERBs being at ankle height, the cuff was positioned on the ankle of the participant. In order for the position to be subject specific it was allowed to rest on the lateral and medial malleoli. In addition, horizontal alignment was ensured by measuring the distance between the floor surface and the ERB attachment point on the ankle cuff as well as the ERB attachment point on the static fixture.

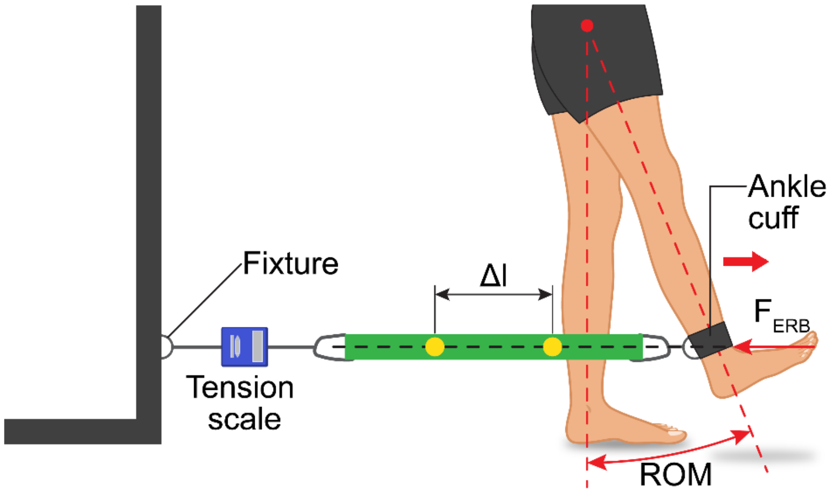


Figure 6: Schematic example of the Experimental set-up showing how fixture, tension scale, ERB and ankle cuff were combined. The yellow dots on the ERB represent the reflective markers use to determine the ERB elongation (Δl).

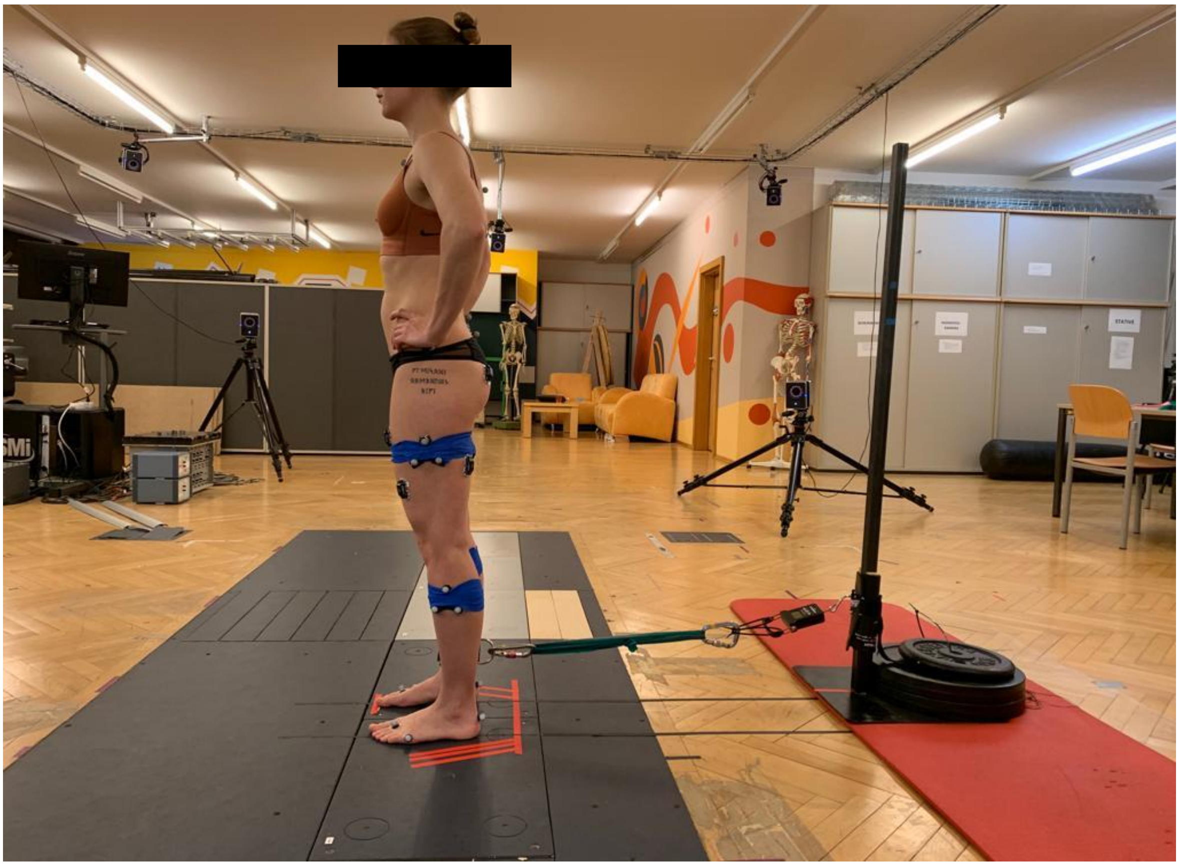


Figure 7: A photo of the flexion trial experimental set-up illustrated by one of the participants. In addition, the measures to prevent the fixation from shifting, namely anti-slip mats and additional weights, can be seen. Furthermore, the floor markings on the force plate can also be seen.

In order to measure the elongation of the ERB during trials, two retroreflective surface markers were placed +10 cm and -10 cm from the centre. A further two markers were then placed on the posterior side of the fixture as well as the anterior of the ankle cuff in order to actually determine dynamic total ERB length during the trials as well as force application points of the ERB resistance load. In order to ensure that ERB starting tension was always the same at the beginning of each exercise trial, a tension scale was fitted between ERB and static fixture. This allowed for the initial starting tension to be defined at 1 kg (9.81 N). The tolerance chosen for the starting tension was ±0.1 kg (0.98 N).

The experimental procedure was divided into three parts (1) static calibration, (2) gait trials and (3) the exercise trials.

As mentioned earlier, the ERB attachment method to the ankle was resolved using an ankle cuff. However, this was not initially clear. Two specific methods were taken into consideration a) looping the ERB behind the ankle, which being a more likely method used in clinical practice as it required no additional equipment, and b) affixing the ERBS using an ankle cuff, which would have the advantage of being secured in place with no possibility of unwanted displacement during exercise execution. The problem was that it was not apparent whether the use of an ankle cuff would significantly change the force produced during the trials or not. To this end a rough calculation was conducted using SOLIDWORKS (Dassault Systèmes, France).

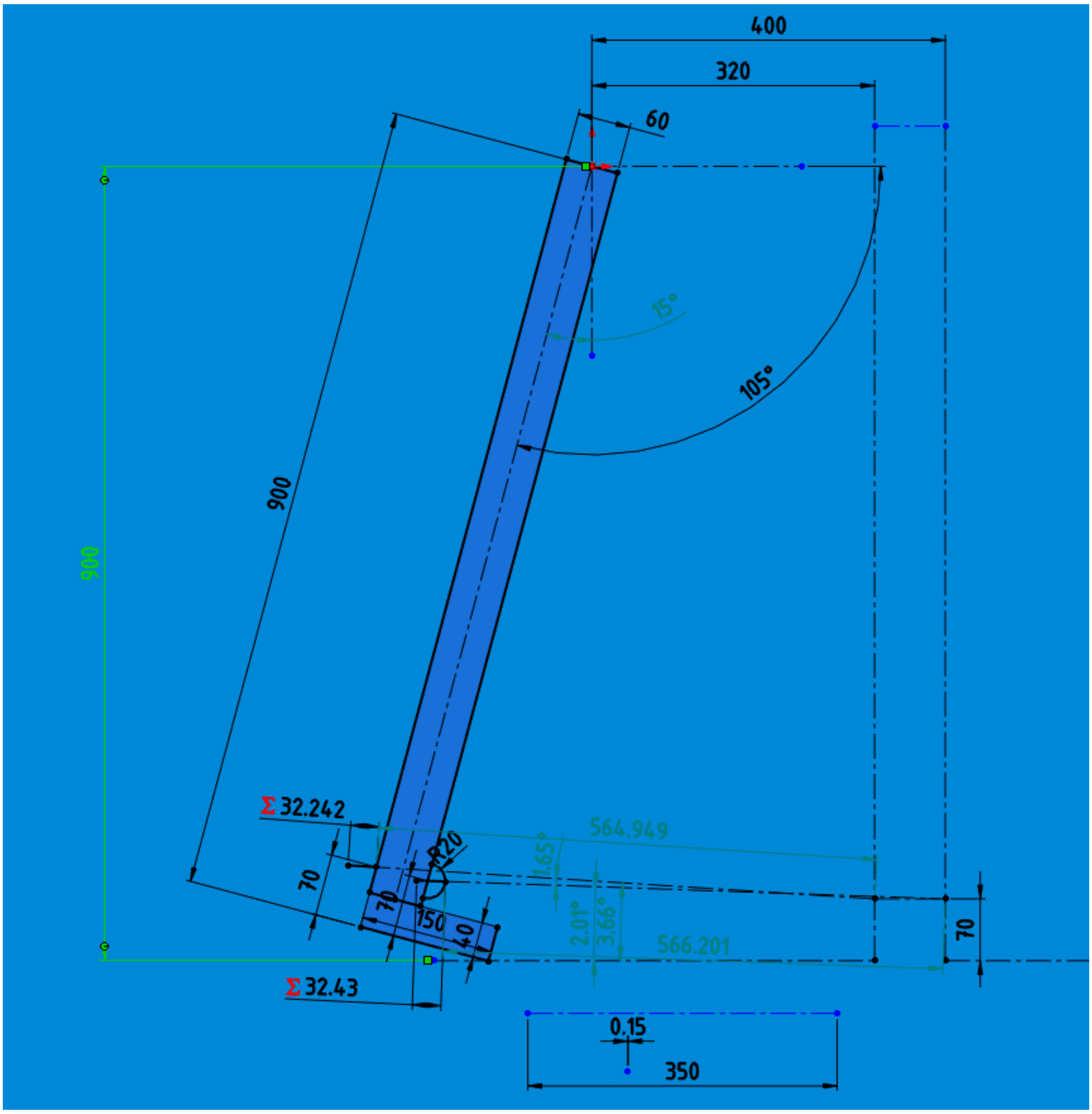


Figure 8: Schematic calculation of the resulting ERB forces acting on the ankle joint during the extension movement using SOLIDWORKS

Approximate calculation of resulting ERB force acting on the ankle joint during the extension movement:

With equal starting length

At $\alpha = 15°$

$$F_{loop} = 32.242\ N\ , \qquad whereby\ \ F_{cuff} = 32.43\ N$$

$$\Rightarrow\ \Delta F = 0.188\ N$$

$$\Delta F_{\%} = \frac{F_{cuff}}{F_{loop}}\ x\ 100\% = 0.58\%$$

At $\alpha = 45°$

$$F_{loop} = 92.25\ N\ , \qquad whereby\ F_{cuff} = 93.12\ N$$

$$\Rightarrow\ \Delta F = 0.87\ N$$

$$\Delta F_{\%} = \frac{F_{cuff}}{F_{loop}}\ x\ 100\% = 0.94\%$$

$$\Delta F_{\%} = \frac{F_{cuff}}{F_{loop}}\ x\ 100\% = 0.94\%$$

α … hip extension angle

F_{loop} … ERB force production using the looping method

F_{cuff} … ERB force production using an ankle cuff

ΔF … Difference in force production between the two methods

$\Delta F_{\%}$ … Difference in force production between the two methods in percent

The approximate calculation shows that even at an extension of the hip of $45°$ the difference in force production in the ERB would only amount to 0.94%. This deviance was deemed acceptable in order to ensure the ERB was secure during exercise execution.

2.6.1 Static Calibration

In order to carry out the marker labelling in the subsequent part of the data collection, a static VICON recording of the participants had to take place beforehand. This enabled a simplified labelling procedure of the gait trials and the exercise trials as well as the insertion of the MSK model in OpenSim in the subsequent part of the data evaluation. During the static recording, the participants were asked to stand with one foot on the left and one on the right of one of the two force plates in a so-called T-pose, whereby care had to be taken to ensure that all relevant markers were visible and that the subject did not move during the static measurement.

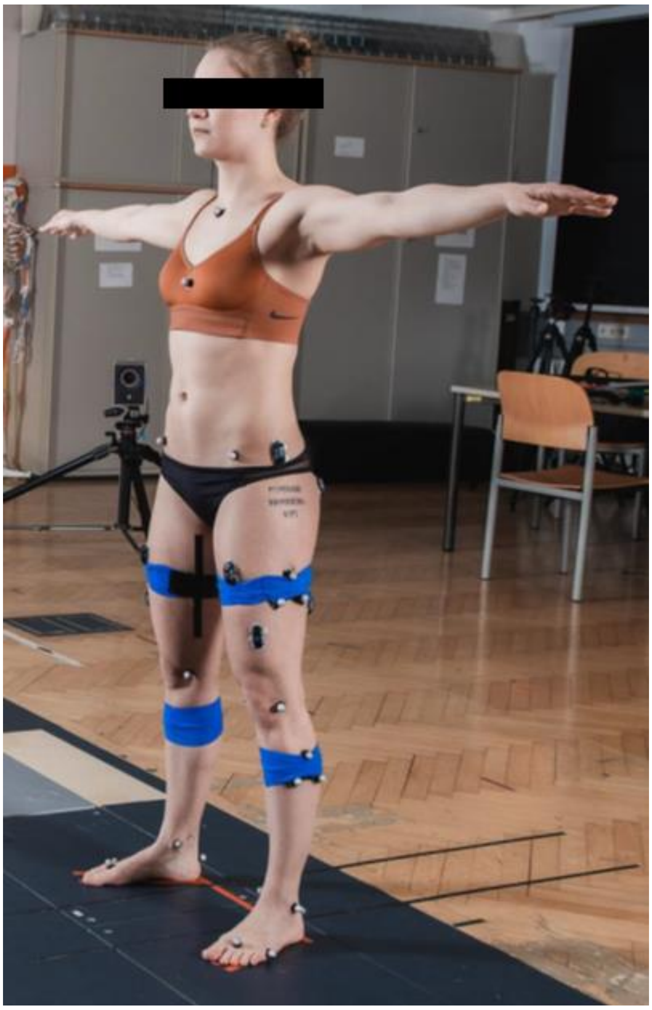


Figure 9: Example of a participant fitted with all 21 retroreflective surface markers, five tri-lateral marker clusters and a wireless EMG electrode, assuming the static T-pose.

2.6.2 Gait Trials

Subsequently, the participants were asked to walk over a distance of approximately 10 m with embedded force plates, at a self-selected gait speed that was as natural as possible. This was carried out until 5 valid trials were recorded from each of the left and right legs. Trials were valid if, with a natural gait, the test person stood with one foot in the middle of one of the force plates embedded in the floor. It was important that everything between initial contact and terminal stance of the gait cycle was detected by the force plate. In addition, the trial was only valid if all markers, preferably over the entire gait cycle, were detected by the VICON system. Gaps, i.e. the momentary non-recognition of the reflective markers by the VICON system, are common in such recordings and can be post-processed. The condition is that the gaps are not too large and that enough other markers are visible during the frame in which the marker is missing in order to extrapolate its position. This is also relevant in the context of the exercise trials.

Figure 10: Example of a participant performing a gait trial while appropriately impacting the force plates with the left and right foot.

2.6.3 Exercise Trials

The exercise trials were conducted after the gait trials. Before the exercises were carried out and recorded, the participants were given time to familiarise themselves with the individual movements. Moreover, the execution of the exercises was closely observed by the instructor and corrected if necessary. In addition, all exercises were performed in two further variations for each resistance type. All exercises were performed in a slower and a faster variation. A metronome was used to standardise the speed of the exercise and the length of the exercise cycle. Participants were instructed to start the movement with a beat, to be at the end of their range of motion by the following beat, and to be back in the initial starting position by the third beat, so that the movement cycle in both cases had the length of 3 beats of the metronome. This resulted in a duration of 3 and 2 s for the slow and fast versions of the movement, respectively. For the slow trials the metronome was set to 40 beats per minute (BPM) and for the faster trials a setting of 60 BPM was chosen. Participants were also asked to keep their hands at their hips and look straight ahead during each exercise. Moreover, care was taken to ensure that the upper body of the participant remained as upright as possible during the exercise. As the speed and duration of the exercise cycle was standardised, the range of motion (ROM) during the exercise was self-selectable by the subjects.

Table 4: Exercise variations and conditions performed by each participant for five repetitions each.

Exercise	Condition	Velocity
No exercise	Walking	Self-selected speed
Squat	No ERB	Slow (3 s)
		Fast (2 s)
	Red ERB	Slow (3 s)
		Fast (2 s)
	Green ERB	Slow (3 s)
		Fast (2 s)
Hip abduction	No ERB	Slow (3 s)
		Fast (2 s)
	Red ERB	Slow (3 s)
		Fast (2 s)
	Green ERB	Slow (3 s)
		Fast (2 s)
Hip flexion	No ERB	Slow (3 s)
		Fast (2 s)
	Red ERB	Slow (3 s)
		Fast (2 s)
	Green ERB	Slow (3 s)
		Fast (2 s)
Hip extension	No ERB	Slow (3 s)
		Fast (2 s)
	Red ERB	Slow (3 s)
		Fast (2 s)
	Green ERB	Slow (3 s)
		Fast (2 s)

The participants stood with each foot on a force plate in the starting position for all exercises. Prior to the start of the trials, and in order to standardise the foot position for each participant, the distance between the anatomical landmarks of the left and right anterior superior iliac spine was measured and subsequently marked on the floor. This had the advantage that the stance width was also patient-specific. To ensure that both feet were parallel to the floor markings (in case of a squat) or to each other (in case of the single leg movements), participants were instructed to place their heels on the markings and, during the single leg movements, point their toes forward.

Figure 11: Floor markings on the force plates used to standardize foot positioning in regards to foot angle and stance width

When performing the squat, the recommendations of Lorenzetti et al. (2018) regarding optimal execution were taken into account. In view of this, an angle of 20 was marked to ensure equal external rotation of the foot during the squat. The patients were also asked to choose the squat depth so that at least a 90 angle was achieved between the upper and lower leg. Before the start of the test series, it was checked whether this minimum execution depth could be realistically performed by all subjects. The subjects were instructed to keep their backs as straight as possible. However, this was not possible for some subjects due to limited flexibility of the dorsal flexion of the foot.

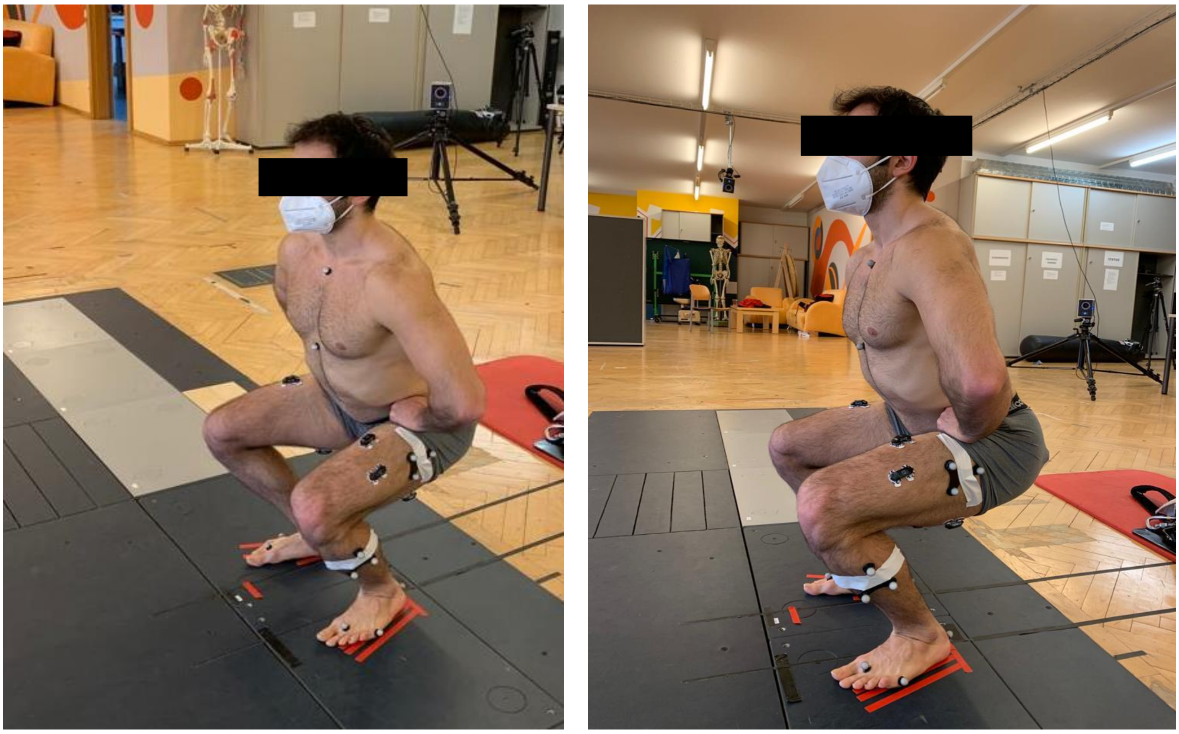


Figure 12: Example of a participant executing a non-ERB loaded body weight squat.

In addition, participants were then asked to keep their knees locked and to maintain a moderate dorsiflexion of the foot throughout the movement cycle. At least five trials were recorded for each exercise condition, e.g. performing the hip flexion exercise using the stiff ERB at the slow pace. An instruction was given to each participant to perform the exercises at a rate of perceived exertion (RPE) of 4 out of 10, corresponding to a contraction intensity of ~40% of maximal voluntary contraction or a training intensity level of a warm-up (Morishita et al., 2013). To avoid fatigue, the subjects were asked to perform each exercise for 5 repetitions. Due to the time needed for the experimental set up to be changed for the next exercise, sufficient rest was ensured between each exercise type, therefore ensuring that progressive fatigue would not misrepresent the data.

In order to keep the movements as similar as possible within the various trials, they were standardised. In general, for all exercises the subjects were asked to place their hands on their hips and instructed to keep the upper body as straight as possible during exercise execution. If the supervisor noticed that the upper body would tilt during the exercise despite the instructions, the subjects were asked to shorten their ROM until they no longer had to compensate for the lack of flexibility, which was especially the case in the abduction, flexion and extension exercises.

Furthermore, data was collected using the right leg for movement execution as well as the left leg. Considering all execution variations, i.e. exercise type, load variation, execution speed and executing leg, and the selected number of repetitions per variation, the number of trials per participant was 250.

2.7 Electromyography

To measure the muscle activity during the exercises, an electromyograph system with wireless electrodes from the company Delsys was used (Delsys, trigno Wireless EMG System). During execution, a total of 12 different muscles of the hip, lower leg and thigh, each of the left and right leg, were measured. The muscles measured by the EMG included: rectus femoris, vastus lateralis, gluteus maximus and medius, biceps femoris and semitendinosus. These were each recorded from the supporting leg and movement leg simultaneously. The location of the attachment was made according to the recommendations of the Seniam group, which can be seen in table 5.

To ensure valid EMG signals, a number of precautions recommended by the Seniam group were considered. First, care was taken to ensure that the electrodes were placed parallel to the muscle fibres at the recommended sensor location. In addition, before attaching the wireless sensors, the skin at each sensor site was appropriately prepared. This included removing body hair and wiping and cleaning the skin with alcohol to remove excess oils from the skin. This in turn ensured good electrode-skin contact, which in turn increased the chance of better EMG recordings, fewer artefacts, and less unwanted noise. All EMG recording were conducted at rate of 1000 Hz.

While EMG data was recorded, it was not used in the publication presented below as its integration into the study would have gone beyond the scope of the project. Nevertheless, the data was included for the sake of completeness in order to expand the data set for possible further investigations.

Table 5: EMG electrode placements according to the recommendation of the SENIAM project

Marker_ID		Side of body	Muscle	Placement
1	R_rect_fem	Right	Quadriceps Femoris; rectus femoris	50% on the line from the anterior spina iliaca superior to the superior part of the patella
2	R_Vast_lat		Quadriceps Femoris; vastus lateralis	2/3 on the line from the anterior spina iliaca superior to the lateral side of the patella.
3	R_Glut_Med		Gluteus Medius	50% on the line from the crista iliaca to the trochanter.
4	R_Glut_max		Gluteus Maximus	50% on the line between the sacral vertebrae and the greater trochanter. This position corresponds with the greatest prominence of the middle of the buttocks well above the visible bulge of the greater trochanter.
5	R_bic_fem		Biceps femoris	50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia.
6	R_semitend		Semitendinosus	50% on the line between the ischial tuberosity and the medial epycondyle of the tibia.
7	L_rect_fem	Left	Quadriceps Femoris; rectus femoris	See R_rect_fem
8	L_Vast_lat		Quadriceps Femoris; vastus lateralis	See R_Vast_lat

8	L_Glut_Med	Gluteus Medius	See R_Glut_Med
10	L_Glut_max	Gluteus Maximus	See R_Glut_max
11	L_bic_fem	Biceps femoris	See R_bic_fem
12	L_semitend	Semitendinosus	See R_semitend

2.8 Three-dimensional motion capture

In order for the joint and limb positioning during the movements to be recorded a Vicon MX13/40 three-dimensional motion capture system (Vicon Motion Systems, Oxford, UK) was used. The system used in the biomechanical motion analysis laboratory of the University of Vienna consisted of a 12-camera optoelectronic infrared system. After initial calibration of each camera the participants were fitted with 21 retroreflective surface markers as well as five trilateral marker clusters. The trilateral marker clusters consisted of three markers in an isosceles triangular arrangement. They were affixed to the participants such as to have one marker in cranial orientation and the other two caudal.

Figure 13: Example of a trilateral marker cluster and its fixture method

Furthermore, six further retroreflective surface markers were fastened to the other components of the experimental setup. Four of the six used were attached to the current ERB in use while, during flexion and extension, one was affixed to the static structure used to anchor the ERB as well as to the ankle cuff worn by the participant. During the squat and abduction movements two markers were attached to the lateral sides of the ERB which was located slightly above the knee.

A number of marks were only necessary for the static calibration and were removed after a valid T-pose was captured. This was possible as the location of specific anatomical landmarks were only required for later use during scaling of the MSK model, while others were required in order for the movement to be tracked. Furthermore, while some of the markers needed to be placed in precise locations, such as on the spinous process of 10th thoracic vertebra, the precise positioning of other markers, namely the trilateral marker clusters, was not necessary and allowed for a certain amount of variance.

All measurements using the motion capture system were taken at a sampling frequency of 100 Hz. The kinematic data was synchronized with the ground reaction force data captured by the two force plates (Kistler Instrumente AG, Switzerland) imbedded into runway. The force plates used had a sample rate of 1,000 Hz. After collection, the marker trajectories were labelled, filtered, and cropped using Nexus 2.11.0 (Vicon Motion Systems, Oxford, UK).

Table 6: Marker set used during motion capture. Grey shaded markers high-light cluster markers.

Segment	Cluster	Marker	Placement
Torso		C7	Spinous process of 7th cervical vertebra
		T10	Spinous process of 10th thoracic vertebra
		Clav	Centered between articuli sterno-clavicularis
		Sternum	Xiphoid process of the sternum
		Right_Back	Infraspinatus [b)]
Pelvis	Pelvis Cluster	PCL_CRAN PCL_RECHTS PCL_LINKS	Resulting center of gravity of the isosceles triangle on center between left and right posterior superior iliac spine
		RASI / LASI [a)]	Right / Left anterior superior iliac spine
Thigh	Right / Left Upper Leg Cluster	R/LCL_UL_CRAN R/LCL_UL_POST R/LCL_UL_ANT	Mid-point between trochanter major and epicondyles lateralis [b)]
		Right/Left_Knee_LAT [a)]	Epicondyles lateralis
		Right/Left_Knee_MED [a)]	Epicondyles medialis
Lower Leg	Right / Left Lower Leg Cluster	R/LCL_LL_CRAN R/LCL_LL_POST R/LCL_LL_ANT	Mid-point between epicondyles lateralis and malleolus lateralis [b)]
		Right/Left_Ankle_LAT [a)]	Malleolus lateralis
		Right/Left_Ankle_MED [a)]	Malleolus medialis
Foot		Right/Left_Heel	Heel leveled with Right/Left_Toe
		Right/Left_Toe	2nd proximal interphalangeal joint
		Right/Left_M5	5th metatarsal head
Elastic Resistance Band		Inside_R/L	+ 10 cm / - 10 cm resp. from band mid-point
		Outside_R/L	Lateral ERB apex during abduction, cuff and fix-point during flexion and extension

Note: a) Remove after static calibration (for calibration purposes only), b) Precise positioning not necessary

2.9 Marker Labelling

After motion capture, the recorded trial need to undergo a marker labelling process. To this end a so-called labelling skeleton template (VST) needed to be created. The VST contains vital information required by the modelling and processing software, in this case Nexus 2.11.0 (Vicon Motion Systems, Oxford, UK), such as information pertaining to the relationship between body segments and joints as well as information regarding the markers themselves, such as maker identification and individual marker placement on the various segments. This provides the software with a generic model that can be scaled to subject specific anthropometry. This means that the process of creating a VST is only necessary once and can subsequently be used for all subjects, as long as the subjects are using the same physical marker set that is described in the VST-file.

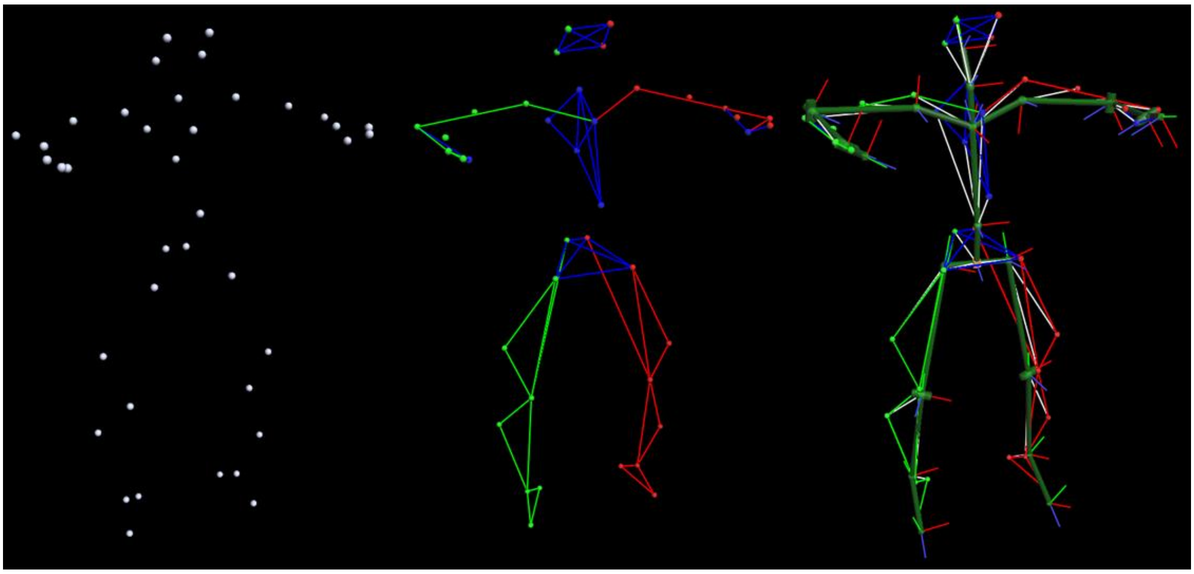


Figure 14: Representation of the three components of motion capture post processing. Figure on the left showing the captured physical marker positioning during a trial, figure in the middle showing the VST, figure on the right showing the VSK. Figure taken from the “Calibrate a labelling skeleton” tutorial (Vicon Motion Systems Limited, 2022).

Following the creation of a VST a labelling skeleton (VSK) can be generated. The VSK specifies the relationship between the physical markers recorded during experimental trials and the aforementioned skeletal model thus creating a subject specific scaled model. This process is repeated for all subjects. The Nexus software has the function to automatically label markers in subsequent trials for the same subject having once been given information regarding the VSK. This means that once a calibration skeleton has been labelled, in the case of the data set described above, the labelling of a static T-pose, other trials would not have to be manually labelled as Nexus is able to extrapolate marker placement data to reconstruct the trial specific VSK using information from the calibrated skeleton. However, in practice this is not always error-free as the software could wrongly label a marker or so-called “gaps” can occur. Gaps are periods in the reconstructed VSK in which specific markers are either missing or cannot be found by the software. The frequency of gap occurrence is largely dependent on data quality. In turn, the data quality varies due to a variety of factors such as a) number of cameras used during motion capture, b) the placement of the cameras used, i.e. is each marker visible by at least once camera at all times, c) movement speed of the subject during capture and d) the frequency of marker occlusion. If such gaps do occur they need to be manually filled which, depending on the length and number of trials, can be extremely time consuming.

While the problem of gaps can often be fixed by manually labelling the marker, the problem of missing markers is more problematic. It is possible that due to camera placement line of sight to the markers is obstructed and certain markers are not visible in some frames. This problem can also be manually rectified providing the gap is not too large and enough other markers are correctly found, providing enough reference points to extrapolate the missing marker position. If, however, this is not the case and the marker cannot be reconstructed using the available data, the trial is unusable and must be discarded.

Due to the size of the data set, which as mentioned consisted of 4,000 individual movement trials, only the movements necessary for study 1 were labelled. This meant that the squat movement as well as all exercises performed with the left leg were not labelled. Thus the number of trials needed to be processed was reduced from the initial 4,000 to a more manageable 1,520 individual movement trials.

As can be seen in Table 7 most exercises exhibit well above a 90% success rate, meaning that they were able to be labelled and subsequently used in the following MSK simulations. Unfortunately, some success rate of some movements and specific participants were slightly lower. For instance, the slow and fast abduction movements using body weight had the lowest exercise trial-success rate with only 78.8% and 80.0% trial success respectively. Similarly, participants P1 and P12 had the lowest trial success rate with 86.3% and 85.3% of all trials recorded being usable. Overall 95.2% of all recorded exercise trial were successfully labelled and used in subsequent MSK simulation for the presented study.

Table 7: Trial-success detailed for each of the 16 participants

	Participant	P1	P2	P3	P4	P5	P6	P7	P8	P9	P10	P11	P12	P13	P14	P15	P16	Trial-Success per exercise [%]
	ERB Calibration pre	x	x	x	x	x	x	x	x	x	x	x	x	x	x	x	x	
	ERB Calibration post	x	x	x	x	x	x	x	x	x	x	x	x	x	x	x	x	
	Static Calibration	x	x	x	x	x	x	x	x	x	x	x	x	x	x	x	x	
Body weight	Gait trails	5	5	5	5	4	5	5	4	5	5	5	5	5	5	5	5	97.5
	Abduction	5	5	5	5	5	5	5	5	5	0	5	1	0	5	5	2	78.8
	Fast Abduction	5	5	5	5	5	5	5	5	5	0	5	0	0	5	4	5	80.0
	Flexion	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
	Fast Flexion	5	5	5	5	5	5	5	5	5	5	5	4	5	5	5	5	98.8
	Extension	5	5	5	5	5	5	5	5	5	5	5	4	5	5	5	5	98.8
	Fast Extension	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
Stiffer ERB	Abduction	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
	Fast Abduction	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
	Flexion	5	0	5	5	5	5	5	5	5	5	5	5	5	5	5	5	93.8
	Fast Flexion	5	0	5	5	5	5	5	5	5	5	5	4	5	5	5	5	92.5
	Extension	2	5	5	5	5	5	0	5	5	5	5	3	4	5	5	5	86.3
	Fast Extension	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
Softer ERB	Abduction	0	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	93.8
	Fast Abduction	0	5	5	5	5	5	5	5	5	5	5	5	3	5	5	5	91.3
	Flexion	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
	Fast Flexion	5	4	5	5	5	5	5	5	5	5	5	5	5	5	5	5	98.8
	Extension	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	5	100
	Fast Extension	5	5	5	5	4	5	5	5	5	5	5	5	5	5	5	5	98.78
	Trial-Success per participant [%]	86.3	88.4	100	100	97.9	100	94.7	99.0	100	89.5	100	85.3	86.3	100	99.0	96.8	**Trial-Success overall [%] 95.2**

2.10 MSK Simulations

MSK simulations were used to calculate the JCF. In the study presented below an inverse approach was used in order to calculate both kinematics, i.e. inverse kinematics (IK), and dynamics, i.e. inverse dynamics (ID). Both the VSK data from the motion capture and the ground reaction forces recorded in parallel with the force plates were used as experimental input data for the simulations.

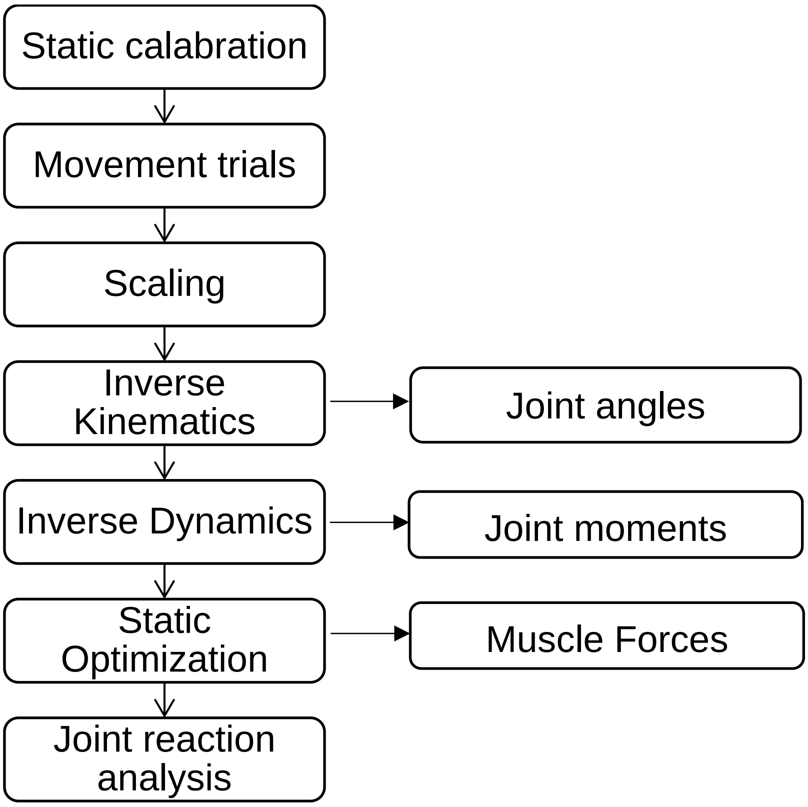


Figure 15: Schematic representation of the inverse MSK modelling approach used in this thesis

As a base model in OpenSim the generic “gait2392” model was used (Delp et al., 1990). The model has 23 degrees of freedom (DoF) and consists of 92 musculotendon actuators to represent 76 muscles in the lower extremities and torso. As a default setting the unscaled model represents a subject that is about 1.8 m tall and has a mass of 75.16 kg. For its use in the presented study the model was scaled to subject specific anthropometry by using surface marker locations at anatomical landmarks and joint centers (Kainz et al., 2017). Where necessary, in case of insufficient markers, certain joints of the model were locked, such as the metatarsophalangeal joints.

In order to perform the MSK simulation, or as a first step to obtain the JCF, the MSK models need to be scaled to fit the MSK models to the participants' anatomy. A measurement-based scaling approach was used to adjust the MSK models according to the marker positions recorded in the motion capture. To achieve this, weightings were assigned to the individual markers. Markers whose positions are known with a high degree of confidence, e.g. anatomical markers such as markers of the right and left anterior superior iliac spine (RASI/LASI), were given a high weighting. Markers whose position are not precisely known are given a low or no weighting, e.g. cluster markers. Those markers that are given a low weighting are more likely to be shifted during the generation of the MSK model than those that have been given a high weighting. Good scaling is achieved when there is a high correspondence between experimental markers, i.e. those captured during the motion capture process, and model markers. This is of particular importance as scaling has a large impact on simulation results (Koller et al., 2021). In general makers weighting are assigned depending on predicted marker confidence. Makers which are assigned a lower confidence due to the higher likelihood of them being displaced during capture, e.g. due to skin motion artifacts or similar, are assigned a lower weighting as a result. Similarly, markers with higher confidence are also assigned a higher weighting. It is suggested that the maximal marker error stay below 2 cm with a Root mean square (RMS) error of below 1 cm. In the study presented all markers of the lower legs were weighted equally while markers close to knee and ankle rotation axis were excluded. All other markers weighting can be seen in the Table below (see Table 8). All scaling errors, as well as Inverse Kinematics simulation errors, were below the best practice recommendations of OpenSim (Hicks et al., 2015).

Table 8: Motion analysis markers and corresponding weights for the inverse kinematics calculations

marker	weight for inverse kinematic	marker name
Sternum	1	sternum
C7	1	7th cervical vertebra
T10	1	10th thoracic vertebra
Clav	1	clavicular notch
Right_Back	0 (not used)	
RASI	0 (not used)	right anterior superior iliac spine
LASI	0 (not used)	left anterior superior iliac spine
PCL_CRAN	10	pelvis cluster - cranial
PCL_LINKS	10	pelvis cluster - left
PCL_RECHTS	10	pelvis cluster - right
LCL_UL_ANT	10	left cluster - upper leg - anterior
LCL_UL_CRAN	10	left cluster - upper leg - cranial
LCL_UL_POST	10	left cluster - upper leg - posterior
Left_Knee_LAT	0 (not used)	left lateral femoral epicondyle
Left_Knee_MED	0 (not used)	left medial femoral epicondyle
LCL_LL_ANT	10	left cluster - lower leg - anterior
LCL_LL_CRAN	10	left cluster - lower leg - cranial
LCL_LL_POST	10	left cluster - lower leg - posterior
Left_Ankle_LAT	0 (not used)	left lateral ankle
Left_Ankle_MED	0 (not used)	left medial ankle
Left_Heel	10	left heel
Left_M5	10	left head of the 5th metatarsal
Left_Toe	10	left toe
RCL_UL_ANT	10	right cluster - upper leg - anterior
RCL_UL_CRAN	10	right cluster - upper leg - cranial
RCL_UL_POST	10	right cluster - upper leg - posterior
Right_Knee_LAT	0 (not used)	right lateral femoral epicondyle
Right_Knee_MED	0 (not used)	right medial femoral epicondyle
RCL_LL_ANT	10	right cluster - lower leg - anterior
RCL_LL_CRAN	10	right cluster - lower leg - cranial
RCL_LL_POST	10	right cluster - lower leg - posterior
Right_Ankle_LAT	0 (not used)	right left lateral ankle
Right_Ankle_MED	0 (not used)	right left medial ankle
Right_Heel	10	right heel
Right_M5	10	right head of the 5th metatarsal
Right_Toe	10	right toe

Furthermore, not only do anthropometrics need to be subject specifically scaled but so does maximum muscle force. While there are multiple approaches in order to scale muscle force, such as with the use of a hand-held-dynamometer (HHD), the study presented used a mass based scaling method (Kainz et al., 2018; van der Krogt et al., 2016). This allowed the maximum isometric muscle forces to be approximated using the following equation:

$$F_{scaled} = F_{generic} * \left(\frac{m_{scaled}}{m_{generic}}\right)^{\frac{2}{3}} \tag{3}$$

F_{scaled} … Scaled muscle force using the mass based scaling method

$F_{generic}$ …Muscle force of the generic gait2392 model

m_{scaled} … Mass of the participant

$m_{generic}$ …Mass to the gait2392 model

Subsequently, the IK was carried out. IK is used to determine the kinematics of the recorded movements. In IK, coordinates and joint angles of the scaled model are changed to achieve the best possible match between the experiment marker, i. e. the VSK, and the model marker, in this case those of the subject specific gait2392 model.

To achieve this, OpenSim uses the principle of Weighted Least Square Minimisation for marker error and coordinate error, whereby coordinate error were not considered in the current thesis. The distance between model marker and experiment marker is kept as small as possible. This is done by solving a weighted quadratic optimisation problem with the aim of minimising marker error, whereby marker error is defined as the distance between an experimental marker and the corresponding virtual marker. The marker weights specify how much this marker error term should be minimised during the solution of the least squares problem. Subsequently, results are squared so that positive and negative values have no influence.

OpenSim uses the data provided to solve for a vector of generalized coordinates, in this case being joint angles, expressed as q, that minimizes the weighted sum of marker errors (Hamner et al., n.d.). In case that only marker error is used the equation of Weighted Least Squares Minimization is a follows:

$$\min_q \left[\sum_{i \in markers} w_i \, \| x_i{}^{exp} - x_i(q) \|^2 \right] \tag{4}$$

q … joint angles

$x_i{}^{exp}$ … position of experimental marker i

$x_i(q)$ … position of the corresponding virtual marker i (which depends on q)

w_i … weight associated with marker i

To reduce marker errors in post-processing, all markers near joint axes were excluded and only cluster markers were tracked during inverse kinematics.

Subsequent to the IK calculations, the inverse dynamics (ID) was performed. With the help of the data determined in the previous steps, such as joint kinematics, external forces, e.g. ground reaction forces, and a subject-specific scaled model, generalized forces such as net forces and torques, were calculated in each joint in any given moment during motion.

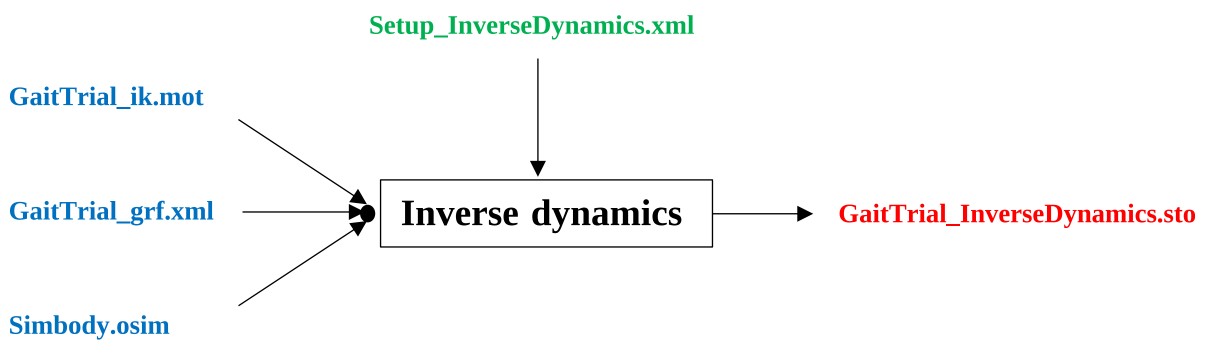
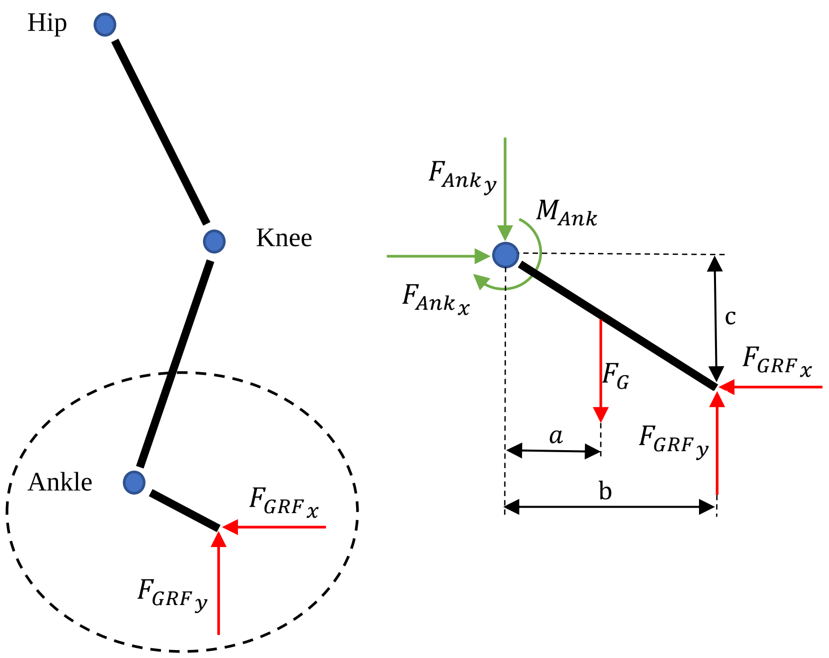


Figure 16: Schematic representation of the required inputs for the inverse dynamics calculation in OpenSim.

This is achieved by means of several equations of motion and the mass-dependent relationship between force and acceleration, i.e. $F = m \, x \, a$. The calculations are carried out from distal to proximal, i.e. the dynamics of the ankle are determined first, then the knee, then the hip, etc. can be derived.



$$\sum F_x = 0 \qquad \sum F_y = 0 \qquad \sum M = 0 \tag{5}$$

F_x … Forces in x-axis

F_y … Forces in y-axis

M … Moment acting on the joint

Figure 17: Example of distal to proximal calculation of forces and moments by means of schematic representation of a leg (simplified to a planar model)

Static optimisation (SO) is used to calculate muscle forces and to determine muscle activation. The reason for optimisation is that the human body has more muscles than DoF, which leads to a very high degree of redundancy. In short, SO used MSK geometry and assumptions pertaining to force distribution in order to estimate the muscle force generated by individual skeletal muscles.

In SO, an attempt is made to determine the optimal muscle force at a given time in the individual muscle that (1) on the one hand generates the net joint moments at a discrete time, (2) does not violate the muscle force limits and (3) optimises the chosen performance criterion.

The aim of a performance criterion (PC) is to capture the goal of the neuro control system. The reason for a PC is based on the muscle force distribution problem, namely, if several muscles are being used during a movement, with which activation was each muscle innervated? Each muscle can contract between 0% and 100%, so the question is how much each muscle is working. Since there are infinite possible solutions to this problem, even if the maximum contraction force is known, the problem of choosing a PC is necessary to solve it.

The difficulty, however, is to define a good PC and to validate that criterion when possible. Examples of possible PCs as mentioned by Zargham et al. (2019) are (1) muscle strength, (2) muscle load / metabolic energy and (3) minimizing the sum of squared muscle activations.

$$f(F_m) = \sum_{m=1}^{nm} F_m \qquad \text{Muscle force}$$

$$f(F_m) = \sum_{m=1}^{nm} \left(\frac{F_m}{PCSA_m}\right)^3 \qquad \text{(Muscle stress)}^3 \qquad (4)$$

$$f(F_m) = \sum_{m=1}^{nm} \left(\frac{F_m}{PCSA_m}\right)^2 \approx \sum_{m=1}^{nm} (a_m)^2 \qquad \text{(Muscle activation)}^2$$

F_m … Muscle force of *m*th muscle

a_m … Parameter that is associated to the muscle properties of the *m*th muscle

$PCSA_m$ … Physiological cross-sectional area of the muscle of the *m*th muscle

In terms of the present study, minimisation of the sum of squared muscle activations was chosen as the PC, as this is also the most commonly used method. Possible validation options are to make a qualitative comparison with experimental EMG or to make comparisons with measured forces, e.g. by instrumented hip implant. Subsequently, the HJCF were determined using the analyse tool and then subjected to further statistical evaluation using MATLAB R2020a (Mathworks Inc., Natick, MA, USA)

The external forces used during the work flow included the ground reaction forces that were recorded via the embedded force plates as well as the external loading precipitated by the elongation of the ERB. The force generated by the different ERBs was plotted using the experimental data from the elongation experiments to create ERB-specific force-elongation curves. These curves were converted to an external force file and applied to the simulations so that the appropriate force was applied depending on the current strain of the ERB used.

2.11 Validation

To ensure that the results of the OpenSim simulations were realistic, it was determined that a validation process was necessary. To accomplish this, two options were considered: (1) a qualitative comparison of the muscle activity generated by the simulations with that measured by the recorded trials using EMG, or (2) a qualitative, visual comparison of the JCF resulting from the simulation with that measured during the same movements using in vivo instrumental hip implants. Despite the availability of EMG data, the second option was chosen as it was deemed a more practical method of validation. For this purpose, the simulated HJCF were compared with those from a public database called OthroLoad (Bergmann, 2008) as can be seen in Figure 18.

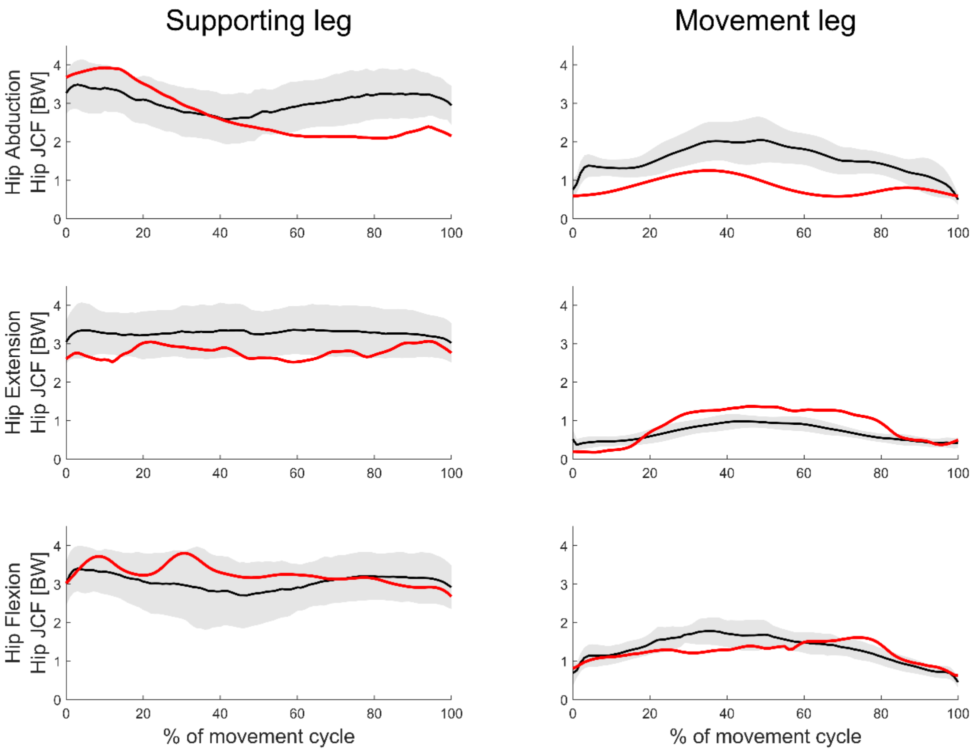


Figure 18: HJCF obtained with the instrumented implant (red waveforms, obtained from participant 'ebl' from the Orthoload database) and the mean (±SD) waveforms from our participant (black waveforms and grey shaded areas) for the three exercises performed without the elastic resistance band.

2.12 Statistical Analysis

For analysis purposes four main parameters were used: (1) Peak Muscle Force, (2) Total Muscle Force, (3) Peak HJCF and Time Force Integral (FTI).

In order to make comparisons of the resulting waveforms, the five relevant trials were taken from all subjects and used to generate a mean value. If one or more of the five trials were discarded in a subsequent step for quality reasons, they were ignored. To enable further analysis, all parameters were time-normalised according to a meaningful criterion. In regard to the gait trials, they were time normalised to one gait cycle. The exercise trials, on the other hand, were normalised according to their movement cycle, whereby the cycle was defined as beginning with the lifting of the movement leg and ending with the touchdown of the movement leg. Both the beginning and the end of the movement cycles were measured using the built-in force plates. To assess inter-subject variability, peak muscle force, total muscle force and peak HJCF were normalised to the body weight of the respective participant.

The FTI was used as an approximation for muscle work as the full calculation of the actual work done by individual muscles during the movements was deemed to be out of scope of the investigation. Although the parameter does not represent the true muscle work, it does give insight into the force profile of a given exercise.

It could be argued that integrating the force over time would provide a predictable result, since the execution speed of the fast and slow trials was controlled by changing the duration of the execution. This would suggest that the longer trials, in this case the slow velocity trials, would automatically yield a larger FTI due to the extra second of execution time. However, the comparisons were made regardless as curve deflections including curve peaks could cause FTI to have a different result.

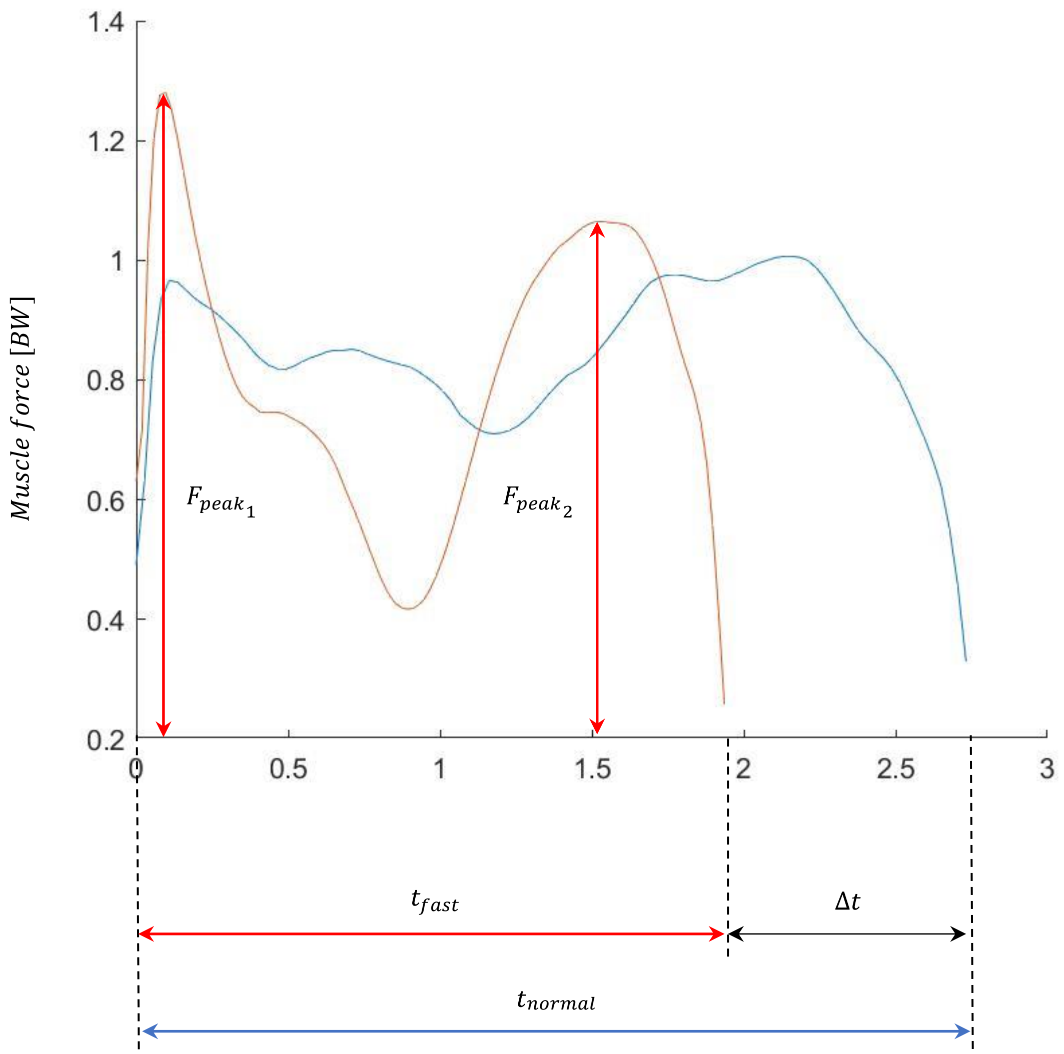


Figure 19: Illustration of the FTI problem by using a force-time diagram of one of the recorded movements. The blue and red curves represent a slow and a fast execution of the same movement respectively.

For the comparison of the resulting waveforms Statistical Parametric Mapping (SPM) was used. SPM is an analytical method used to enable statistical analysis of continuous, usually biomechanical, data (Honert & Pataky, 2021). Formally used in the analysis of brain imaging data sequences, SPM has found more use in the field of biomechanics in recent years (Honert & Pataky, 2021). The advantage of SPM is that, unlike discrete statistical tests which only examine individual values within a continuous time curve while ignoring the majority of the data, SPM allows for continuous time analysis, which makes for a much more illustrative and comprehensive comparison. In short, the main difference with SPM is that while it operates very similarly to fundamental statistical analyses such as t-tests, ANOVA and linear regression, it extends them to one-dimensional data analysis. In the presented study, the SPM was further used to provide a binary result of statistical significance for individual points and curve regions within the identified movement cycles. To achieve curve alignment, all waveforms were aligned after the common event of the movement leg lifting off, meaning that the force measured by the force plate associated to the movement leg fell to 0 N. To perform the SPM analysis, the SPM1D package for Matlab from http://www.spm1d.org/ was taken and applied to the relevant data (Pataky, 2010).

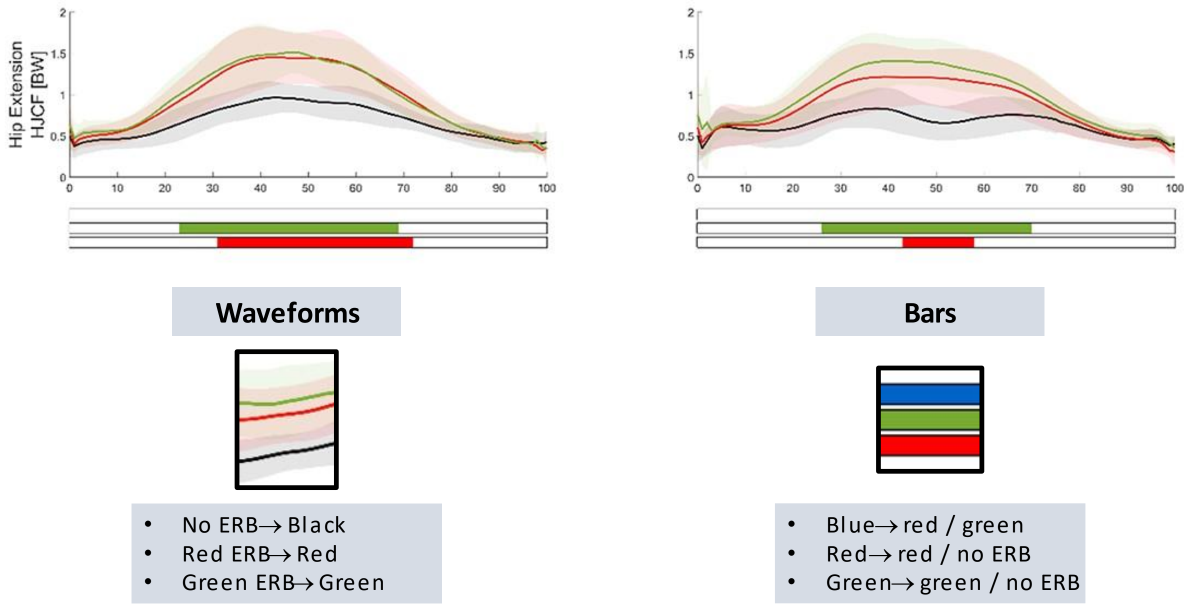


Figure 20: Example of SPM analysis showing mean (±SD) muscle force waveforms measured in the movement leg during hip extension exercises during the slow (left subplots) and fast (right subplots) velocities trials. Green, red, and black waveforms represent the stiffer, softer, and no ERB, respectively. Colored bars beneath each plot indicate significant differences between waveforms, whereas the green, red, and blue (first) bars represent significant differences between the stiffer vs. no ERB, softer vs. no ERB, and stiffer vs. softer ERB, respectively.

While SPM was used for the first hypothesis, classical statistical methods were used to investigate the other two hypotheses. More information is provided in the presented study.

2.13 Research objectives and questions

The aim is to gain new insights into the loading behaviour of hip OA rehabilitation exercises though analysis and characterisation of the HJCF of typical rehabilitation exercises for hip OA.. Based on the knowledge gained through this study, professionals in the field of physiotherapy, rehabilitation and prevention will be able to make more informed and precise recommendations for rehabilitation movements for hip OA.

Research question: To what extent does the HJCF of typical rehabilitation exercises for hip OA calculated using MSK simulations differ from each other and from those of a normal gait cycle?

Hypothesis: The peak HJCF values of the rehabilitation exercises are significantly lower than those of a typical gait cycle.

Aim: Quantifying the different muscle forces and joint loading during resistance band exercises.

Hypothesis:

(1) muscle forces and JCF are higher when using a stiffer (green) resistance band compared to a softer (red) resistance band and no resistance band,

(2) movement execution with a higher velocity will increase peak hip JCF but decrease total muscle forces

(3) peak and total muscle forces but not peak hip JCF of the executing leg will be higher compared to walking

3 Publication P1

Quantifying Muscle Forces and Joint Loading During Hip Exercises Performed With and Without an Elastic Resistance Band

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Abstract

An increase in hip joint contact forces (HJCFs) is one of the main contributing mechanical causes of hip joint pathologies, such as hip osteoarthritis, and its progression. The strengthening of the surrounding muscles of the joint is a way to increase joint stability, which results in the reduction of HJCF. Most of the exercise recommendations are based on expert opinions instead of evidence-based facts. This study aimed to quantify muscle forces and joint loading during rehabilitative exercises using an elastic resistance band (ERB). Hip exercise movements of 16 healthy volunteers were recorded using a three-dimensional motion capture system and two force plates. All exercises were performed without and with an ERB and two execution velocities. Hip joint kinematics, kinetics, muscle forces, and HJCF were calculated based on the MSK simulations in OpenSim. Time-normalized waveforms of the different exercise modalities were compared with each other and with reference values found during walking. The results showed that training with an ERB increases both target muscle forces and HJCF. Furthermore, the ERB reduced the hip joint range of motion during the exercises. The type of ERB used (soft vs. stiff ERB) and the execution velocity of the exercise had a minor impact on the peak muscle forces and HJCF. The velocity of exercise execution, however, had an influence on the total required muscle force. Performing hip exercises without an ERB resulted in similar or lower peak HJCF and lower muscle forces than those found during walking. Adding an ERB during hip exercises increased the peak muscle and HJCF but the values remained below those found during walking. Our workflow and findings can be used in conjunction with future studies to support exercise design.

Keywords

Elastic resistance band, MSK simulations, hip joint contact force, muscle force, hip strengthening exercises, OpenSim, rehabilitation

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The author’s contribution

CB and HK conceived the original idea and wrote the paper. CB collected the data and prepared the data for the simulations. WK performed the simulations. WK and FD processed the data. WK, HK, and CB performed statistical data analysis. HK supervised the project.

3.1 Introduction

Persistent symptomatic problems of the hip joint have been shown to cause a substantial impact on the overall health in the older population (Dawson et al., 2005). This is especially problematic considering that one in five people aged 65 years and older experience hip pain (Dawson et al., 2004). Some of the conditions that cause this hip pain, such as osteoarthritis (OA), have no cure and can cause an accelerated progression, leading to a high rate of surgical interventions (Gossec et al., 2005). Joint degeneration in the hip and knee OA is associated with altered gait patterns (Astephen et al., 2008; Eitzen et al., 2012; Meyer et al., 2015, 2018). These altered gait patterns often lead to joint pathomechanics such as high joint contact forces, which accelerate the progression of the disease (Meireles et al., 2017; Richards et al., 2018).

Compensatory movement strategies found in patients with hip OA are often a result of the observed hip muscle weakness (Meyer et al., 2018). A systemic review by Loureiro et al. (2013) highlighted that the affected legs of hip OA show significantly lower muscle strength compared to both the contralateral leg and/or healthy controls. Strengthening the joint supporting muscles is used as a conservative treatment to improve the quality of life of patients and to slow down the progression of OA (Zhang et al., 2008; Nho et al., 2013). The required muscle stimulus for muscle strengthening can be achieved with different exercise modalities (Hofmann et al., 2016; Iversen et al., 2018).

For muscle-strengthening exercises, elastic resistance bands (ERBs) are especially an easy-to-use, cheap, and effective alternative to conventional resistance-training equipment (Cambridge et al., 2012; Sundstrup et al., 2014; Calatayud et al., 2015; Aboodarda et al., 2016). Previous studies investigated the material properties of ERBs (Simoneau et al., 2001; Santos et al., 2009; Uchida et al., 2016). These studies highlighted that the resistance force increases linearly with the elongation of the ERB. Furthermore, the force–elongation characteristics differ between ERBs with different stiffnesses. Due to this predictive, linear behavior, as well as to the other benefits mentioned above, ERBs present an ideal and practical training method for rehabilitation exercises. However, to the best of the knowledge of the authors, no studies assessed the impact of ERBs on muscle and joint contact forces.

Strengthening the hip muscles increases the stability of the joint and reduces joint contact forces (Retchford et al., 2013; Meyer et al., 2018). In other words, increased stability due to a more balanced muscle force distribution reduces femoral head translation and therefore decreases joint contact forces. This is especially critical because the presence of increased hip joint contact forces (HJCFs) is one of the main contributing mechanical causes of hip OA and its progression (Recnik et al., 2009; Felson, 2013). Therefore, the knowledge, understanding, and subsequent control of these forces are essential for building a progressive rehabilitation program. Despite the link between muscle weakness, joint contact forces, and OA progression, recommendations for rehabilitative muscle-strengthening exercises are often based on an expert opinion instead of the supporting scientific evidence (Conaghan et al., 2008; Zhang et al., 2008).

Only a small number of studies investigated the impact of hip exercises on HJCF. While a plethora of literature on the relationship between HJCF and movements, such as walking, running, and stair climbing exist (Heller et al., 2001; Bergmann et al., 2004; Lenaerts et al., 2008; Giarmatzis et al., 2015, 2017; Wesseling et al., 2015; Meyer et al., 2018; Kainz et al., 2020), only sporadic research has been done regarding other activities, such as single-leg standing or cycling (Bergmann et al., 2001; Varady et al., 2015; Damm et al., 2017) and even less that have dealt with specific hip-strengthening exercises. Catelli et al. (2020) compared HJCFs during a squat between patients with the cam-type femoroacetabular impingement for both pre- and post-hip-corrective surgeries and with those of a healthy control in which they found no significant difference. In vivo measurements via instrumented endoprosthesis showed that only weight-bearing exercises caused significantly high HJCF (up to 441% of the body weight), whereas most of the others, such as non-weight-bearing, isometric exercises, did not (Schwachmeyer et al., 2013). Investigation on the impact of alternative weight-bearing training modalities, such as ERB exercises, on HJCF, is still missing.

The goal of this study was to (1) quantify the muscle forces and the accompanying loading on the hip joint during ERB exercises, which target muscles shown to promote joint stability, and (2) compare these forces to those observed during walking. Our participants performed hip-strengthening exercises with two different ERBs and execution velocities. During all exercises, the participants were standing on one leg and performed the movement with the contralateral leg. We hypothesized that (1) muscle forces and HJCF are higher when using a stiffer ERB compared with those using a softer ERB and no ERB, (2) movement execution with a higher velocity will increase the peak HJCF but decrease the total muscle forces, and (3) the peak and total muscle forces but not the peak HJCF of the movement leg will be higher compared with walking. In addition, we analyzed and compared joint kinematics, joint kinetics, and ERB forces between the different exercise modalities, i.e., different ERB and execution velocities, to get a comprehensive overview of the impact of ERB exercises on the MSK system.

3.2 Material and Methods

Three-dimensional motion capture data and ground reaction forces were collected during the typical hip muscle-strengthening exercises used in the rehabilitation of hip pathologies. These data were used for MSK simulations to estimate the muscle forces and HJCF.

3.2.1 Participants

Sixteen healthy adults (11 men and 5 women) with no pre-existing or acute lower limb pathologies were recruited *via* word of mouth and participated in our study. Their average ± SD age, weight, height, and body mass index were 27 ± 4 years, 70.7 ± 12.5 kg, 1.75 ± 0.10 m, and 22.9 ± 2.8 kg m^{-2}, respectively. The research ethics and methods of the study were approved by the Ethics Committee of the University of Vienna (00579), and all participants were informed regarding the purpose of the study and gave their written consent before participation.

3.2.2 Exercises

All participants performed rehabilitation exercises that aimed to strengthen the following hip-stabilizing muscles: (a) hip abductors including gluteus medius, gluteus minimus, tensore facie latae, and piriformis (Valente et al., 2013; Meyer et al., 2018); (b) hip flexors including rectus femoris, iliacus, psoas, iliocapsularis, and sartorius (Zhang et al., 2008); and (c) hip extensors including gluteus maximus, biceps femoris, semitendinosus, and semimembranosus (Loureiro et al., 2013). The exercises were performed in a standing straight-legged position, and each muscle group was targeted with a separate exercise. All exercises were first performed without the use of an ERB and then subsequently with two different elastic band types that differed in their resistance-elongation characteristics. The order of the used ERB was the same for each participant.

All exercises were performed at a slow executing speed and a fast executing speed. A metronome was used to standardize the execution velocity. Participants were instructed to start the movement with a beat, to be at the end of their range of motion by the following beat, and to be back in the initial starting position by the third beat. Using 40 beats per minute for the slow and 60 beats per minute for the fast variant resulted in an exercise duration of 3 and 2 s for the slow and fast movement execution, respectively.

Each participant performed five gait trials by walking over a 10 m long designated runway with embedded force plates at a self-selected walking speed. Each gait trial was cropped to one gait cycle. Subsequently, each participant executed the following three exercises: (1) a standing single-leg abduction, (2) a standing single-leg hip extension, and (3) a standing single-leg hip flexion (Figure 1). During all exercise trials, the participants were instructed to keep their hands on their hips and to look straight ahead and use their full range of motion while keeping their torso as still and upright as possible. During the initial position of all exercises, the participants were standing with each foot on one force plate. To standardize the foot position for each participant, the distance between the left and right anterior superior iliac spine anatomical landmarks was measured and marked on the floor prior to the start of the trials. The participants were asked to place their heels on the markings and point their toes forward, to ensure both feet were parallel to each other. Furthermore, the participants were asked to keep their knees straight and to exert a slight dorsiflexion with the foot during the entire course of the movements. For each exercise and condition (Table 1), at least five trials were collected for each condition. Each participant was instructed to perform the exercises at a rate of the perceived exertion of 4 out of 10, which corresponds to a contraction intensity of ~40% of the maximum voluntary contraction or a training intensity level of a warm-up (Morishita et al., 2013).

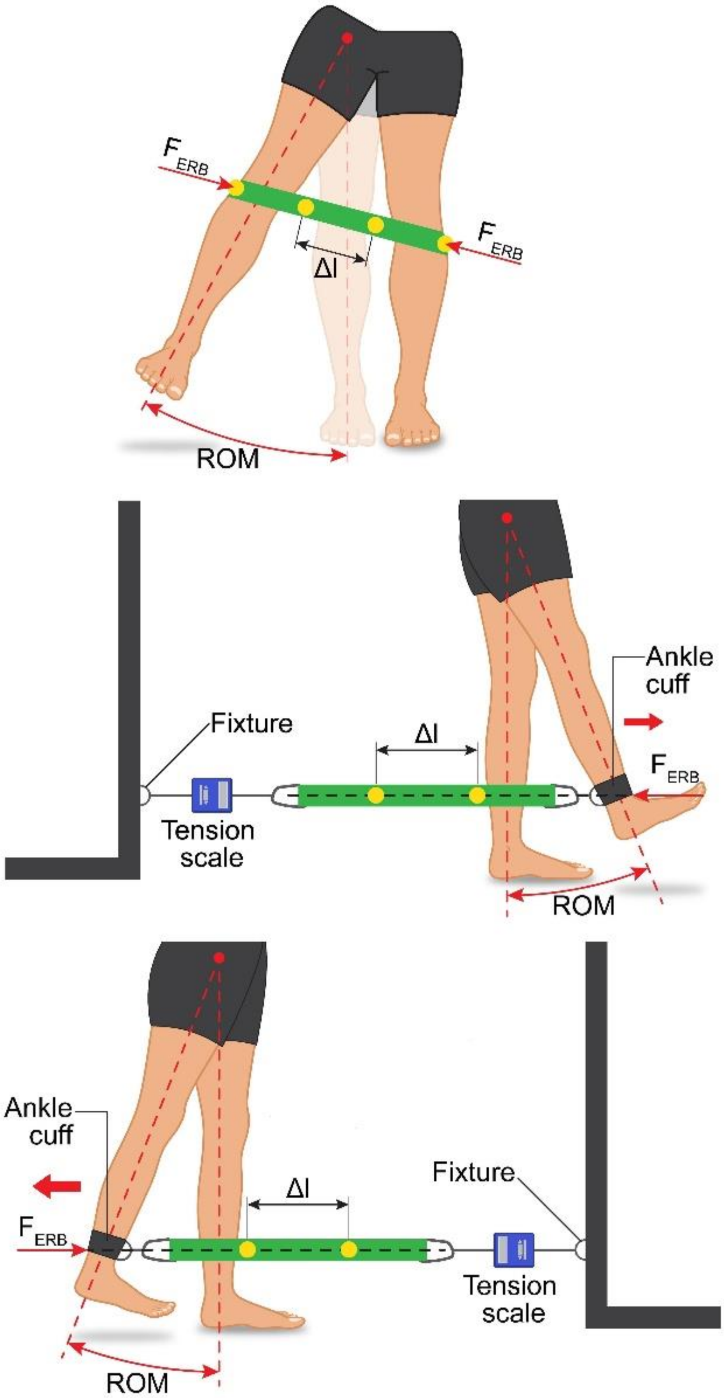

Figure 21. Experimental set-up showing movement execution as well as ERB fastening method and position for all three exercises. (1) standing single-leg abduction, (2) standing single-leg hip extension and (3) standing single-leg hip flexion. The dimension Δl shows the measured marker displacement used to calculate the force production of the ERB (i.e. F_{ERB}).

Table 9: Exercise variations and conditions

Exercise	Condition	Velocity
No exercise	Walking	Self-selected speed
Hip abduction	No ERB	Slow (3 s) Fast (2 s)
	Red ERB	Slow (3 s) Fast (2 s)
	Green ERB	Slow (3 s) Fast (2 s)
Hip flexion	No ERB	Slow (3 s) Fast (2 s)
	Red ERB	Slow (3 s) Fast (2 s)
	Green ERB	Slow (3 s) Fast (2 s)
Hip extension	No ERB	Slow (3 s) Fast (2 s)
	Red ERB	Slow (3 s) Fast (2 s)
	Green ERB	Slow (3 s) Fast (2 s)

For the ERB trials, the bands were secured in place to ensure that they would not move during the exercises. For the flexion and extension trials, a fixture was used that was aligned with the movement leg (Figure 1). An ankle cuff was used to attach the ERB to the leg, while the other end of the ERB was attached to the fixture. Furthermore, a tension scale was inserted between the fixture and the resistance band to standardize the band tension at the beginning of each exercise. A starting tension of 1 kg (9.81 N), with a tolerance of ±0.1 kg (0.98 N), was chosen. The cuff was placed above the ankle and was allowed to sit on the lateral and medial malleoli. To ensure a horizontal alignment of the ERB, the distance between the floor and the ankle cuff joint was measured and the joint on the opposite side between the fixture and the ERB was adjusted to match. The ERB, as well as the ankle cuff and fixture, was fitted with markers. To track the ERB elongation, two markers were placed on the ERB, each +10 cm and −10 cm from the midpoint of the loops, respectively. The two markers placed on the ankle cuff and the fixture were placed on the two lateral ERB loop apices during an abduction. These markers were subsequently used to define the force application point of the ERB.

3.2.3 Three-dimensional motion capture

To capture the movement of our participants, 21 retroreflective surface markers (Table 2) and 5 trilateral marker clusters were attached to the lower body and torso of each participant. In addition, four markers were used to track the ERB elongation. The subsequent marker trajectories were captured using a 12-camera optoelectronic system (Vicon Motion Systems, Oxford, UK) at a sampling frequency of 100 Hz. Simultaneously, synchronized ground reaction forces were collected via two embedded force plates (Kistler Instrumente AG, Switzerland) at a sample rate of 1,000 Hz. After collection, the marker trajectories were labeled, filtered, and cropped using Nexus 2.11.0 (Vicon Motion Systems, Oxford, UK).

Table 10: Marker set used for collecting the movement of our participants. Grey shaded markers highlight cluster markers.

Segment	Cluster	Marker	Placement
Torso		C7	Spinous process of 7th cervical vertebra
		T10	Spinous process of 10th thoracic vertebra
		Clav	Centered between articuli sterno-clavicularis
		Sternum	Xiphoid process of the sternum
		Right_Back	Infraspinatus [b)]
Pelvis	Pelvis Cluster	PCL_CRAN	Resulting center of gravity of the isosceles triangle on center between left and right posterior superior iliac spine
		PCL_RECHTS	
		PCL_LINKS	
		RASI / LASI [a)]	Right / Left anterior superior iliac spine
Thigh	Right / Left Upper Leg Cluster	R/LCL_UL_CRAN	Mid-point between trochanter major and epicondyles lateralis [b)]
		R/LCL_UL_POST	
		R/LCL_UL_ANT	
		Right/Left_Knee_LAT [a)]	Epicondyles lateralis
		Right/Left_Knee_MED [a)]	Epicondyles medialis
Lower Leg	Right / Left Lower Leg Cluster	R/LCL_LL_CRAN	Mid-point between epicondyles lateralis and malleolus lateralis [b)]
		R/LCL_LL_POST	
		R/LCL_LL_ANT	
		Right/Left_Ankle_LAT [a)]	Malleolus lateralis
		Right/Left_Ankle_MED [a)]	Malleolus medialis
Foot		Right/Left_Heel	Heel leveled with Right/Left_Toe
		Right/Left_Toe	2nd proximal interphalangeal joint
		Right/Left_M5	5th metatarsal head
Elastic Resistance Band		Inside_R/L	+ 10 cm / - 10 cm resp. from band mid-point
		Outside_R/L	Lateral ERB apex during abduction, cuff and fix-point during flexion and extension

Note: a) Remove after static calibration (for calibration purposes only), b) Precise positioning not necessary

3.2.4 Elastic resistance band

Two ERBs of the brand Theraband (Thera-Band, OH, USA) were used in this study. The green ERB was the stiffer one, whereas the red ERB was the softer one. These two ERBs were chosen because they are often recommended by physiotherapists for home exercises.

To ensure that the ratio of displacement to force production was consistent between the participants, as well as to validate the assumption of a linear relationship between the force and elongation, both ERBs used were evaluated before and after performing all trials of each participant. To verify the aforementioned assumptions, a series of different weights were affixed to the ERB and the elongation was measured using the Vicon system. The stiffer ERB was loaded with 0, 0.5, 1.0, 2.5, 5, and 7.5 kg, and the softer ERB was loaded with 0.0, 0.5, 1.0, 2.5, and 5 kg. The displacement of the attached reflective markers was measured and was subsequently used to fit a line to the experimental force, elongation data (Figure 2). The equation of the fitted line based on the ERB tests before the dynamic data collection with each participant was used to create the external force file for the dynamic MSK simulations (described in detail below). A paired t-test indicated no significant differences ($p > 0.05$) between the recorded elongation and the obtained fitted lines before and after the collection of the dynamic trial.

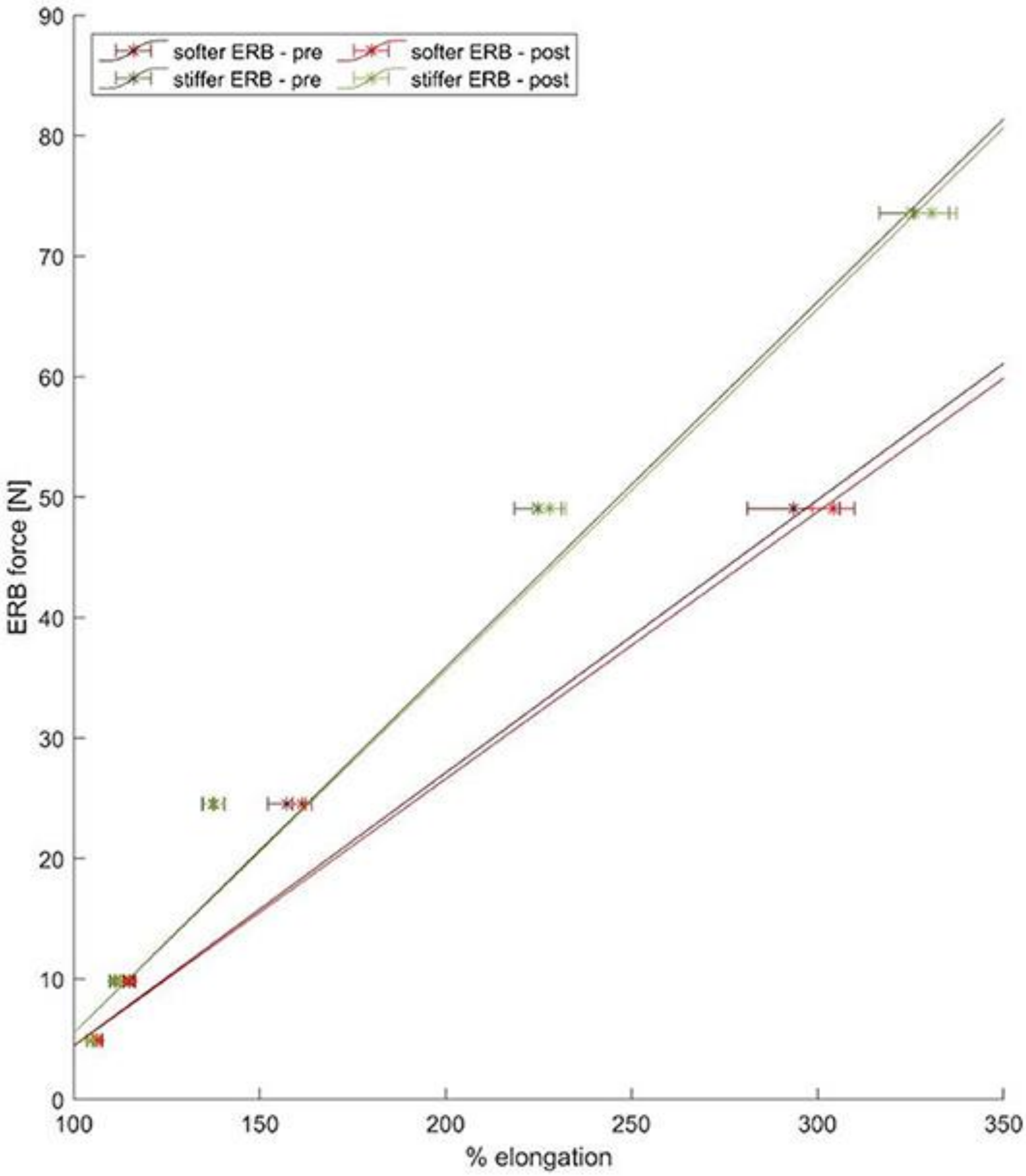

Figure 2. Mean Force-Elongation curve of the green (green line) and red (red line) ERB obtained from the experimental data points (six points for the green and five for the red ERB) based on the pre (blue lines) and post (red lines) data collection validation experiments.

3.3 MSK simulations

The generic "gait2392" OpenSim model (Delp et al., 1990) was scaled to the anthropometry of each participant using surface marker locations at anatomical landmarks and joint centers (Kainz et al., 2017). Due to insufficient markers at the foot, the metatarsophalangeal joints of the models were locked. The maximum isometric muscle forces were scaled depending on the body mass of the participants by Equation (1) (van der Krogt et al., 2016; Kainz et al., 2018).

$$F_{scaled} = F_{generic} * \left(\frac{m_{scaled}}{m_{generic}}\right)^{\frac{2}{3}} \quad (1)$$

The models of the participants and the corresponding motion capture data were used to run inverse kinematics followed by inverse dynamics, static optimization by minimizing the sum of squared muscle activations, and joint reaction load analyses with MATLAB R2020a (Mathworks Inc., Natick, MA, USA) and OpenSim 4.1 (Seth et al., 2018). The external force file used during the OpenSim simulations included the ground reaction forces from the force plates and the ERB forces obtained from the elongation of the ERB and the force–elongation curves. During inverse kinematic, all markers close to joint axes were excluded and only the cluster markers were tracked. Detailed information about which markers were included and their weighting factors can be found in the Supplementary Material. All scaling errors, as well as simulation errors, were below the best practice recommendations of OpenSim (Hicks et al., 2015).

3.4 Validation of simulations

To validate our simulation results, a qualitative visual comparison of the HJCF measured during each exercise was made with those found on OrthoLoad (Bergmann, 2008), a public database of HJCF measured in vivo with instrumented hip implants. The HJCF from all exercises in this study showed a reasonable agreement with the values from OrthoLoad (details can be found in the Supplementary Material).

3.5 Data Processing and Statistical Analysis

For all analyzed parameters, the average waveform from approximately five trials per condition and participant was calculated and time-normalized. Gait trials were normalized to one gait cycle, whereas exercises were normalized to the movement cycle using the force plate data of the movement leg (movement started and ended when the foot left and hit the force plate, respectively). Furthermore, muscle forces and HJCFs were normalized to the body weight of each participant. For our first hypothesis, muscle force and HJCF waveforms were compared between exercises without and with ERBs. For each exercise, only the muscle group of interest was compared between the different conditions (e.g., average hip adductor muscle forces for the hip adductor exercise). For our second hypothesis, the peak HJCF and the force–time integral (FTI) were determined for each condition and compared between the slow and fast exercise executions. The FTI was used to estimate the total amount of muscle force needed for each exercise. We calculated the FTI by integrating the force of the corresponding muscle group over time, e.g., FTI for the hip adductor exercise was calculated by integrating the hip adductor muscle forces over time (Beltman et al., 2004; Ortega et al., 2015). For our third hypothesis, the peak HJCF, FTI, and peak muscle forces of the respective muscle groups of each exercise were compared with the same muscle groups during walking. Statistical parametric mapping (Pataky, 2010) based on the SPM1D package for Matlab (http://www.spm1d.org/) was used to statistically compare the waveforms for our first hypothesis. Within the SPM1D package, two-tailed scalar trajectory t-tests (SPM{t}) with Bonferroni adjusted alpha level (i.e., $p = 0.05/3 = 0.0167$ for the following comparisons: no ERB vs. softer ERB; no ERB vs. stiffer ERB; and softer vs. stiffer ERB) were chosen to compare the muscle forces and HJCF waveforms between exercises with and without ERB. IBM SPSS Statistics, version 27.0. (IBM, New York, USA) using repeated-measures ANOVA with a set significance level of $p < 0.05$ was used to compare the discrete parameters for our second and third hypotheses. For the second hypothesis, we used repeated measures ANOVA with the factors “ERB” (no ERB, softer ERB, stiffer ERB) and “speed” (fast, slow), whereas for our third hypothesis, we used repeated measures ANOVA with the factor “movement” (gait, exercise without ERB, exercise with softer ERB, exercise with stiffer ERB) and contrast-coded post-hoc tests (gait vs. all exercises) in case that the ANOVA revealed significant group differences. Repeated-measure results were verified with Greenhouse–Geisser corrections where the Mauchly test of sphericity determined the heterogeneity of covariance. In case of significant main effect, pairwise post-hoc comparison using Bonferroni-adjusted alpha levels was conducted. In addition, we assessed if there was a significant interaction between ERB and speed.

3.6 Results

All the following figures, tables, and subsequent results presented pertain to the movement leg. The figures and graphs displaying the results of the standing leg and detailed statistical results (i.e., exact p-value for each comparison, F scores, partial eta-squared) can be found in the Supplementary Material of this study.

3.6.1 Study performance

While 16 participants performed the experiments, at various points in the data processing, some trials were either unusable or missing, e.g., missing markers, isolated muscle EMG signals unusable, or not all movement conditions performed. If this was the case, the incomplete or distorted data for the specific trial were discarded. However, this only applied to the isolated trial of the specific parameter. The total number of participants used in the final analysis is shown as “N” in Supplementary Tables 1, 2.

3.6.2 Hypothesis 1: Muscle Forces and HJCF Are Higher When Using a Stiffer ERB Compared to Those Using a Softer ERB and No ERB

In regard to our first hypothesis, we found significantly higher ($p < 0.0167$) muscle forces during the middle part of the movement cycle when using an ERB (soft or stiff) compared to those using no ERB for hip extension and flexion exercises (Figure 3). HJCFs were significantly higher ($p < 0.0167$) during the middle part of the movement cycle when using an ERB (soft or stiff) compared to that using no ERB for hip extension exercises (fast and slow) and the fast hip flexion exercises (Figure 4). Performing the hip exercise with a stiffer or softer ERB did not show any significant differences in muscle forces and HJCFs.

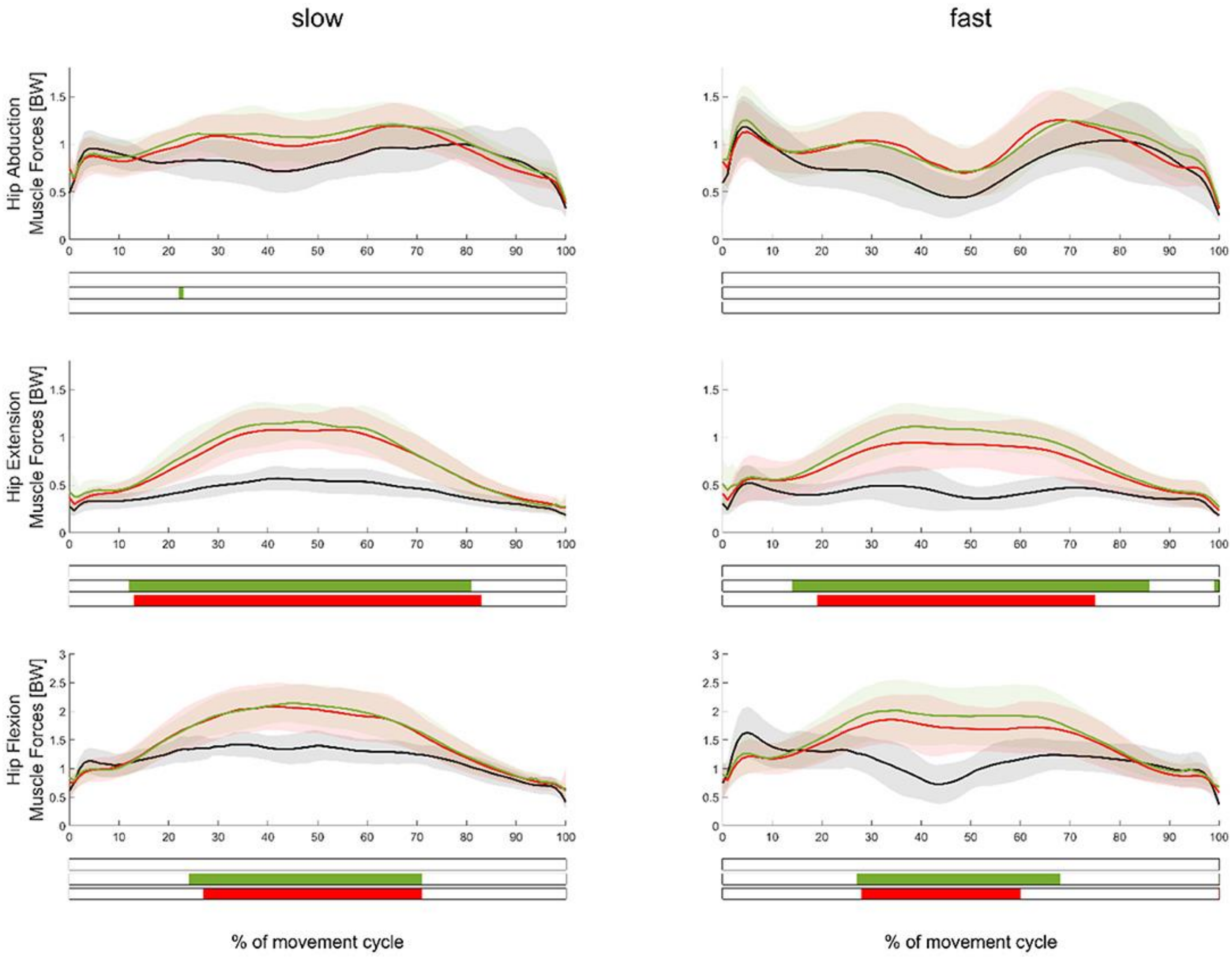

Figure 3. Mean (±SD) muscle force waveforms measured in the movement leg during hip abduction (top), extension (middle), and flexion (bottom) exercises, as well as during slow (left subplots) and fast (right subplots) velocities. Green, red, and black waveforms represent the stiffer, softer, and no ERB, respectively. Colored bars beneath each plot indicate significant differences between waveforms, whereas the green, red, and blue (first) bars represent significant differences between the stiffer vs. no ERB, softer vs. no ERB, and stiffer vs. softer ERB, respectively.

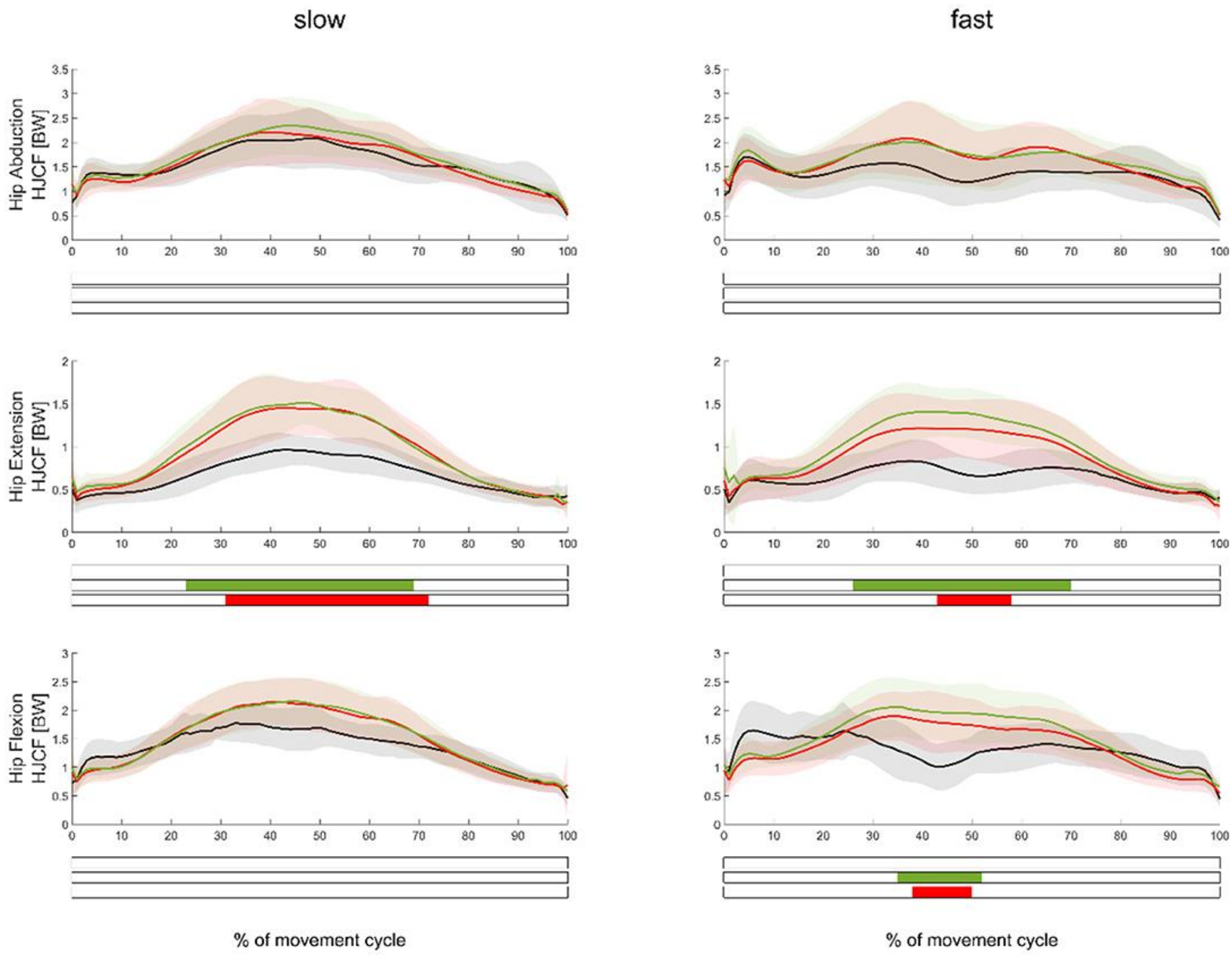

Figure 4. Mean (±SD) HJCF waveforms measured in the movement leg during hip abduction (top), extension (middle), and flexion (bottom) exercises, as well as during slow (left subplots) and fast (right subplots) velocities. Green, red, and black waveforms represent the stiffer, softer, and no ERB, respectively. Colored bars beneath each plot indicate significant differences between waveforms, whereas the green, red, and blue (first) bars represent significant differences between the stiffer vs. no ERB, softer vs. no ERB, and stiffer vs. softer ERB, respectively.

The comparison of joint kinematics between the exercise execution variations without ERB and those with softer and stiffer ERBs showed several significant differences (Figure 5). The use of an ERB significantly decreased ($p < 0.0167$) the range of motion for hip extension and flexion exercises. Joint kinematics between exercises performed with the softer and stiffer ERBs were not significantly different. Similar to our muscle force results, joint moments of hip flexion and extension exercises were significantly higher ($p < 0.0167$) during the middle of the movement cycle when using an ERB compared with the exercise without the ERB (Supplementary Figure 2). ERB forces were only significantly higher when using the stiffer compared with those using the softer ERB (Figure 6) during hip abduction exercises.

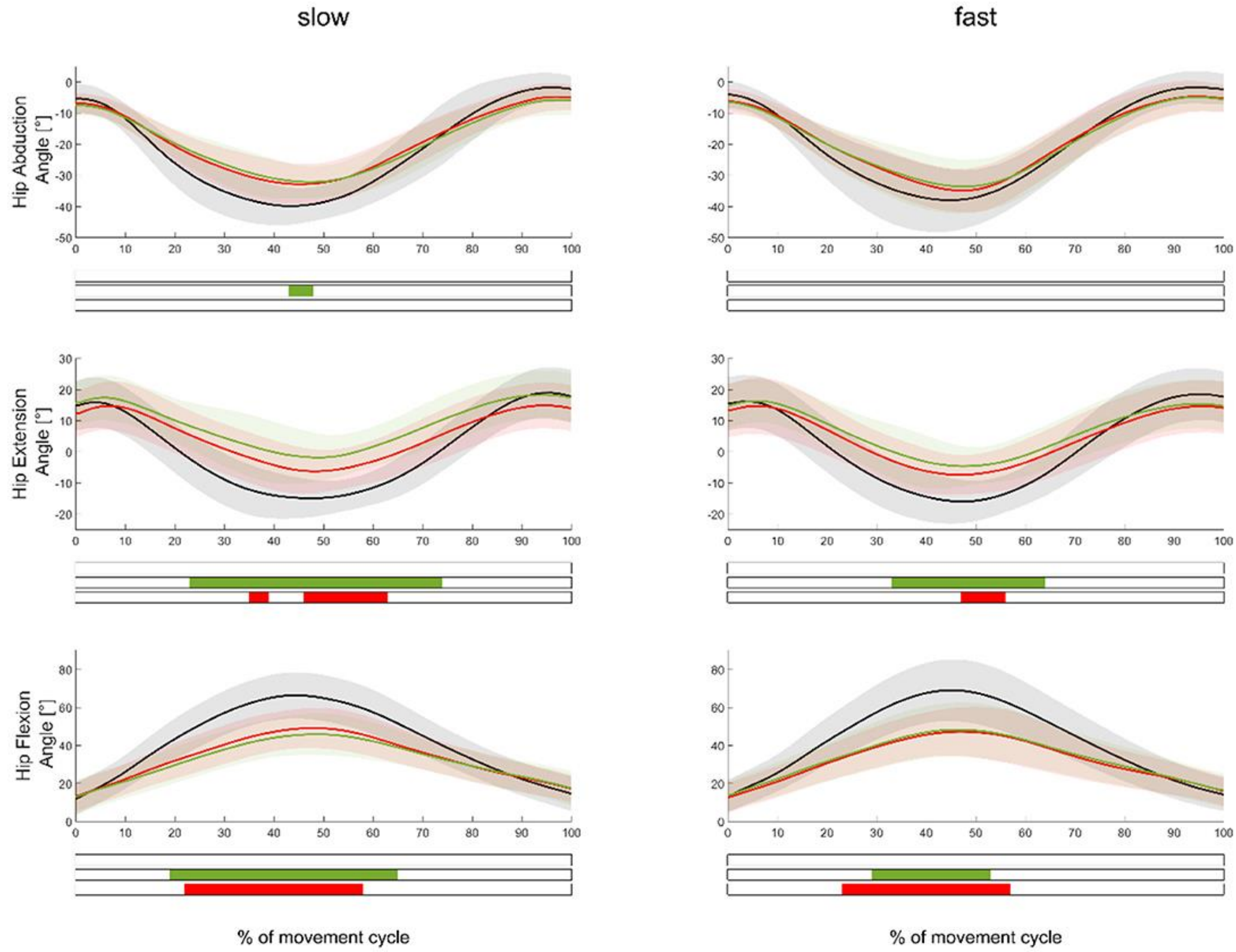

Figure 5. Mean (±SD) hip angle waveforms measured in the movement leg during hip abduction (top), extension (middle), and flexion (bottom) exercises, as well as during slow (left subplots) and fast (right subplots) velocities. Green, red, and black waveforms represent the stiffer, softer, and no ERBs, respectively. Colored bars beneath each plot indicate significant differences between waveforms, whereas the green, red, and blue (first) bars represent significant differences between the stiffer vs. no ERB, softer vs. no ERB, and stiffer vs. softer ERB, respectively.

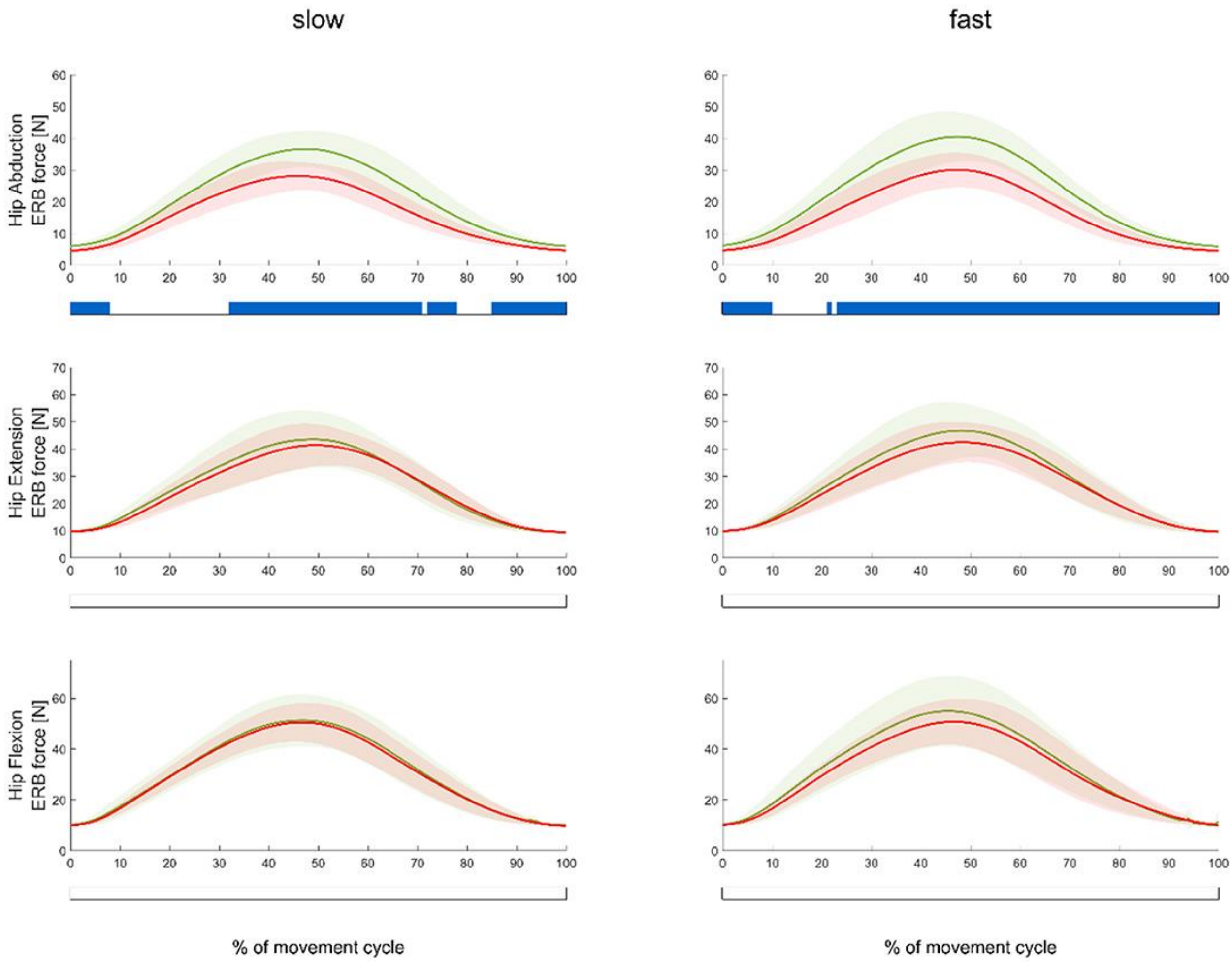

Figure 6. Mean (±SD) ERB forces during slow (left) and fast trials (right) measured in the softer (red waveform) and stiffer (green waveform) ERBs during abduction (top), extension (middle), and flexion (bottom) exercises. Blue bars beneath each plot indicate significant differences between the forces of the softer and stiffer ERBs.

3.6.3 Hypothesis 2: Movement Execution With a Higher Velocity Will Increase the Peak HJCF but Decrease the Total Muscle Forces (FTI)

Independently of the use of an ERB or not, comparing exercises performed with the slow and fast velocities did not show any significant differences ($p = 0.987$) in the peak HJCF. However, consistent with our assumption, the slow velocity trials showed a significantly higher ($p < 0.001$) FTI than those of the fast velocity trials (Figure 7). This was true for all exercises and execution variants. We only found a significant interaction ($p = 0.009$) between ERB and the speed for peak HJCF when performing hip extension exercises. ERB forces were not significantly different between exercises performed with different velocities (refer to the Supplementary Figure 4 in the Supplementary Material).

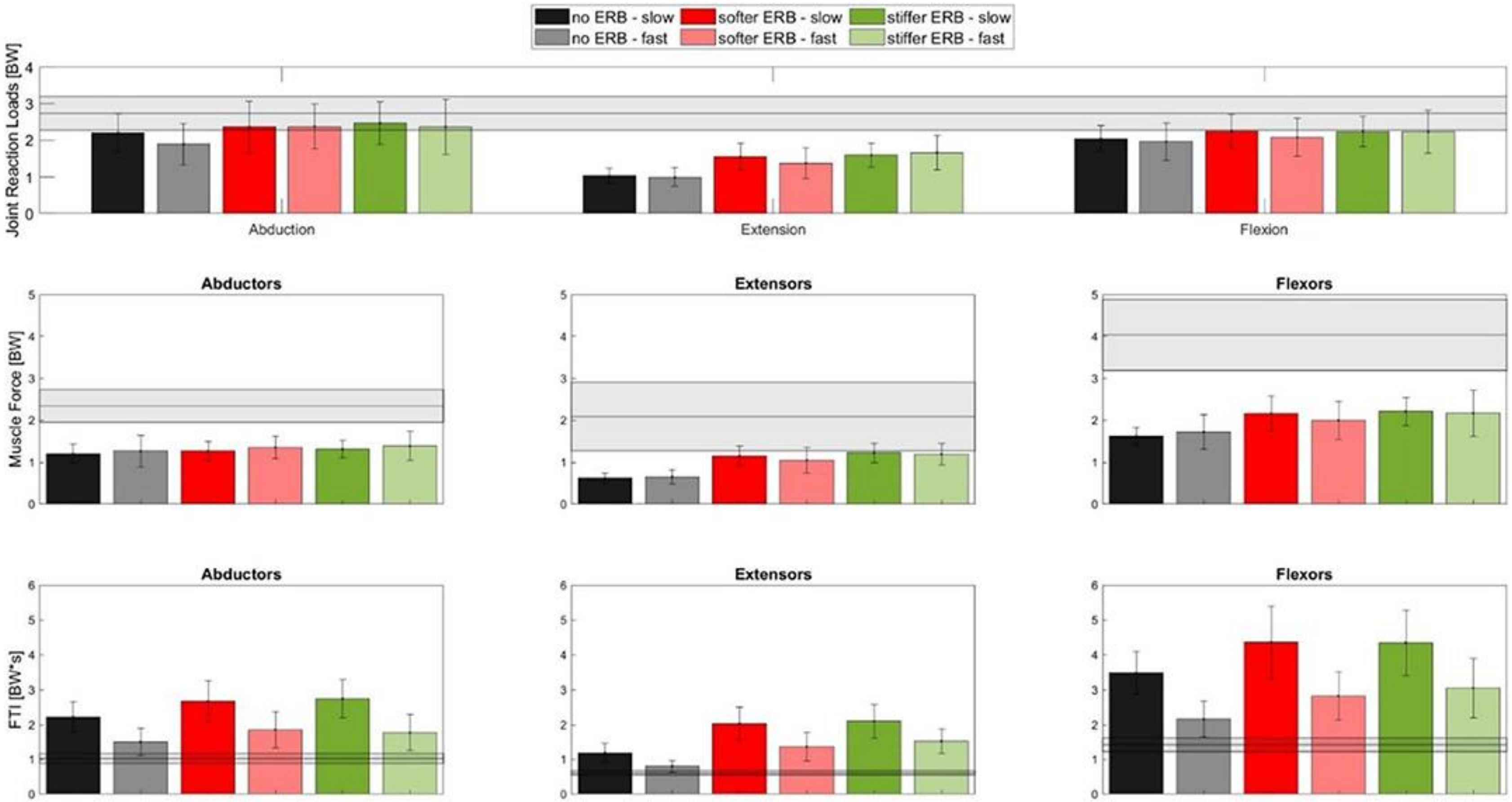

Figure 7. Bar plots showing the mean (±SD) of the peak HJCF (top row) during abduction (left), extension (middle), and flexion (right) as well as the peak muscle forces (middle row) and FTI (bottom row) of the respective target muscle groups (abductors, extensors, and flexors, shown in the left, middle, and right sides, respectively) in the movement leg. Each bar represents one of the execution variants (see the legend above). The gray horizontal bar in every plot depicts the mean (±SD) values of the respective parameter measured during a gait cycle.

3.6.4 Hypothesis 3: Peak and Total Muscle Forces but not Peak Hip JCF of the Movement Leg Will Be Higher Compared to Those During Walking

In all exercises, the peak muscle forces in the movement leg were significantly lower ($p < 0.05$) compared with the respective peak values during walking (Figure 7).The total required muscle forces, i.e., FTI, of each corresponding muscle group of the respective exercise were significantly higher ($p < 0.05$) compared to the same muscle group during the gait trials for all exercises except the hip abduction exercise performed with the fast velocity. Compared to walking, the peak HJCFs were significantly lower ($p < 0.001$) during the fast- and slow-performed hip extension exercises. The peak HJCFs were also significantly lower ($p = 0.017$) compared to walking in the fast-executed hip flexion exercises without an ERB.

3.7 Discussion

The primary aim of this study was to evaluate the muscle forces and associated loads on the hip joint during ERB exercises and to compare these forces with those observed during walking. In agreement with our first hypothesis, both, the stiffer and the softer ERBs, consistently showed significantly higher muscle forces over most of the exercises when compared with those found during exercises performed without an ERB. This outcome confirmed the general assumption that an increase in the training load due to the ERB would lead to higher muscle forces of the targeted musculature. However, comparing muscle forces between the softer and stiffer ERBs did not show a significant difference. HJCF analyses showed a similar trend, with no significant differences in HJCF between the softer and stiffer ERBs. These findings were surprising and partly contradicted our first hypothesis. Comparing the two execution velocities showed, contrary to our second hypothesis, that the variance in velocity does not change the HJCF. However, the required total muscle forces (FTI) were consistently lower during the exercises performed with the fast compared to those with the slow velocity, partly confirming our second hypothesis. When comparing the exercises with walking, the peak muscle forces were significantly lower during all exercises, which was in contrast to our third hypothesis. In addition, the peak HJCFs were similar or significantly lower during the exercises compared with that during walking. On the other hand, the required total muscle forces, i.e., FTI, were significantly higher when exercising with an ERB compared to those during walking, which partly confirmed our third hypothesis.

One of the main goals of the study was to not only quantify the HJCFs but also put them into a perspective using a known and understood metric, which, in our case, were the HJCF found during a gait cycle. However, as walking is generally recommended as a form of aerobic exercise to patients with hip pathologies, such as hip OA, this only gives us a rough idea rather than a full spectrum of acceptable HJCF in people with hip OA (Zhang et al., 2008). This begs the question as to what could be considered to be the upper acceptable limit of HJCF of therapeutic, muscle-strengthening exercises. In people with hip pathologies, jogging is generally considered unsuitable due to the high impacts and the resulting HJCF, which are as high as 5.74 body weight at a speed of 6 km/h (Zhang et al., 2008; Giarmatzis et al., 2015). Taking this into consideration, the HJCFs observed during the ERB exercises in this study were relatively low and did not exceed the values obtained during walking.

Interestingly, compared to the hip flexion and extension exercises, adding an ERB had a minor impact on the muscle and HJCF during the hip abduction exercises. The ERB was attached to the ankle during the hip flexion and extension trials, whereas during the hip abduction trials, the ERB was attached to the femoral condyles. Different ERB locations lead to different moment arms, which might be the reason why adding an ERB barely changed the muscle forces and HJCF during hip abduction exercises.

Comparing the slow- with the fast-performed exercises did not show any significant differences in HJCF. This highlights that a certain variation in execution velocity does not influence hip joint loading and that the velocity of the exercise execution could be determined based on the preference of a patient. Slow velocities, however, significantly increased the total required muscle forces (i.e., FTI) during the exercises compared with fast velocities. In addition to the longer execution duration, slow execution velocities might lead to an increase in agonist–antagonist coactivation due to increased demand on joint stability and therefore a higher FTI. Our simulation results, however, did not confirm this assumption (refer to the Supplementary Material). From a combined training and joint loading perspective, exercises performed with slow velocities are recommended because less repetition and therefore, fewer loading cycles with peak HJCFs are needed to obtain the same FTI compared with the fast-performed exercises.

The magnitude of the HJCF during walking found in this study (mean peak HJCF 2.7 ± 0.45 BW over all participants) was in agreement with the previous findings using instrumented implants (2.4–2.8 BW) but slightly lower compared with the previous simulation studies (3.7–4.9 BW) (Bergmann et al., 1993, 2001; Valente et al., 2013; Modenese et al., 2018; Passmore et al., 2018; Kainz et al., 2020). Different walking velocities, biomechanical models, computational approaches, and study population might be the reason for the observed difference in HJCF between this study and the findings from the previously published simulation studies (Giarmatzis et al., 2015; Kainz et al., 2016; Trinler et al., 2019).

The total required muscle force per exercise (i.e., FTI) increased, as expected, together with an increasing time under tension (slow vs. fast movement execution). The FTI was used as an approximation for muscle work and, although the parameter does not represent the true muscle work, it does give insight into the force profile of a given exercise. Hence, the combination of HJCF, peak muscle forces, and FTI could be used as parameters of exercise control and training design. Furthermore, the ERB type should be chosen to fit the hip range of motion of a patient, as well as to fit the current strength level. The stiffer the ERB, the lesser the range of motion is required to produce the same force. Hence, people with a limited range of motion would potentially benefit from a stiffer ERB to achieve adequate training.

This study included the following limitations. First, we only investigated the impact of two types of ERBs on muscle forces and hip joint loading. The chosen ERBs are often used during rehabilitation exercises but only slightly differed in their force, elongation characteristics. Using different ERBs with larger differences in their force, elongation characteristics (e.g., yellow vs. black ERB from the brand Theraband) would probably lead to more significant differences between the ERBs. Second, greater differences in execution velocities between our slow- and fast-performed trials could lead to different results. These velocities were, however, chosen intentionally as they represent realistic velocities used during rehabilitation exercises. Third, our participants were healthy adults without any known hip pathologies. A different study cohort, e.g., people with hip OA, could perform the exercises with slightly different hip kinematics, which would affect the obtained muscle forces and hip joint loading (Wesseling et al., 2015; Higgs et al., 2019; Diamond et al., 2020). We, however, expect that the relative results, e.g., HJCF due to exercise performed with vs. without an ERB, would be similar to a different study cohort. Fourth, different models and computational approaches might lead to slightly different results (Pieri et al., 2018; Hoang et al., 2019). Fifth, in our ERBs, the relationship between force and elongation was not perfectly linear (Figure 2). Assuming a non-linear relationship and fitting a curve, i.e., second-degree polynomial curve, to our experimental data would have led to a better fit but this would not have affected our findings or conclusion (refer to the Supplementary Material). We chose a linear relationship to be consistent with the previous publications (Hughes et al., 1999). Sixth, considering that a standard gait cycle usually takes around 1 s and our exercise trials took 2 and 3 s for the slow and fast movement executions, respectively, our FTI comparison between the exercises and walking should be interpreted with caution.

3.8 Conclusion

This study highlighted the impact of hip exercises with an ERB on the targeted muscle forces and HJCF. The type of ERB used and the exercise execution velocity had a minor impact on the peak muscle forces and HJCF. Execution velocity, however, does affect the total muscle force required for an exercise. Performing hip exercises without an ERB resulted in similar or lower peak HJCF and lower muscle forces than those found during walking. Adding an ERB during hip exercises increases the peak muscle and HJCF but the values remained below those found during walking. The total muscle forces, i.e., FTI, during hip exercises exceeded the values obtained during walking. This study showed the impact of rehabilitative hip exercises on hip joint loading and the surrounding muscle forces.

3.9 Data Availability Statement

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

3.10 Ethics Statement

The studies involving human participants were reviewed and approved by Ethics Committee of the University of Vienna (00579). The patients/participants provided their written informed consent to participate in this study.

3.11 Conflict of Interest

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

3.12 Publisher's Note

All claims expressed in this article are solely those of the authors and do not necessarily represent those of their affiliated organizations, or those of the publisher, the editors and the reviewers. Any product that may be evaluated in this article, or claim that may be made by its manufacturer, is not guaranteed or endorsed by the publisher.

3.13 Acknowledgments

The authors would like to acknowledge Liam Strasser and David Deimel for their help during data collection and to all participants who took the time to participate in this study. Furthermore, the authors would like to thank Ass-Prof Dr. Peter Gröpel for his support during their statistical analyses.

3.14 Appendix

3.14.1 Validation of our simulations

Mean HJCF waveforms of the respective exercises without the ERB across all participants for the movement leg and the stance (supporting) leg were comparable to those obtained from a participant with an instrumented hip implant from the Orthoload database (Figure S1). In all exercises without the ERB, HJCF were higher in the supporting leg compared to the movement leg in our participants, which was in agreement with the participant from the Orthoload database. The shape and maximum values of the Othoload waveforms were similar to the waveforms obtained from our participants, with exception of the HJCF of the movement leg for the hip abduction exercise, which showed higher values in our participants.

Visual comparison between the HJCF from the participants in our study with those found on the Orthoload database showed a reasonable agreement for all exercises (Figure 3) (Bergmann, 2008). Nevertheless, some differences were evident between the HJCF waveforms, especially for the hip abduction exercise. It should be noted that all the HJCF waveforms from the Orthoload database were from one single participant. Differences in HJCF between our results and the values from Orthoload might be caused by a combination of differences in hip kinematics and movement execution velocities, additionally to the different methods to obtain the HJCF (simulations versus in-vivo measurement).

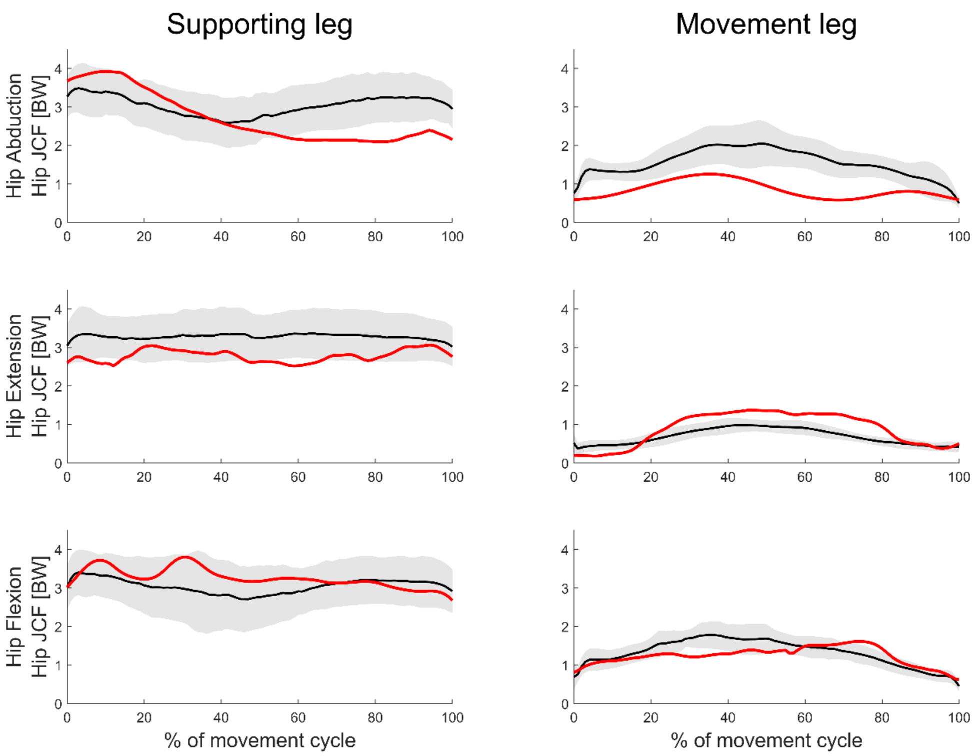

Figure S1. HJCF obtained with the instrumented implant (red waveforms, obtained from participant ‘ebl’ from the Orthoload database) and the mean (±SD) waveforms from our participant (black waveforms and grey shaded areas) for the three exercises performed without the elastic resistance band.

3.14.2 Additional figures/tables

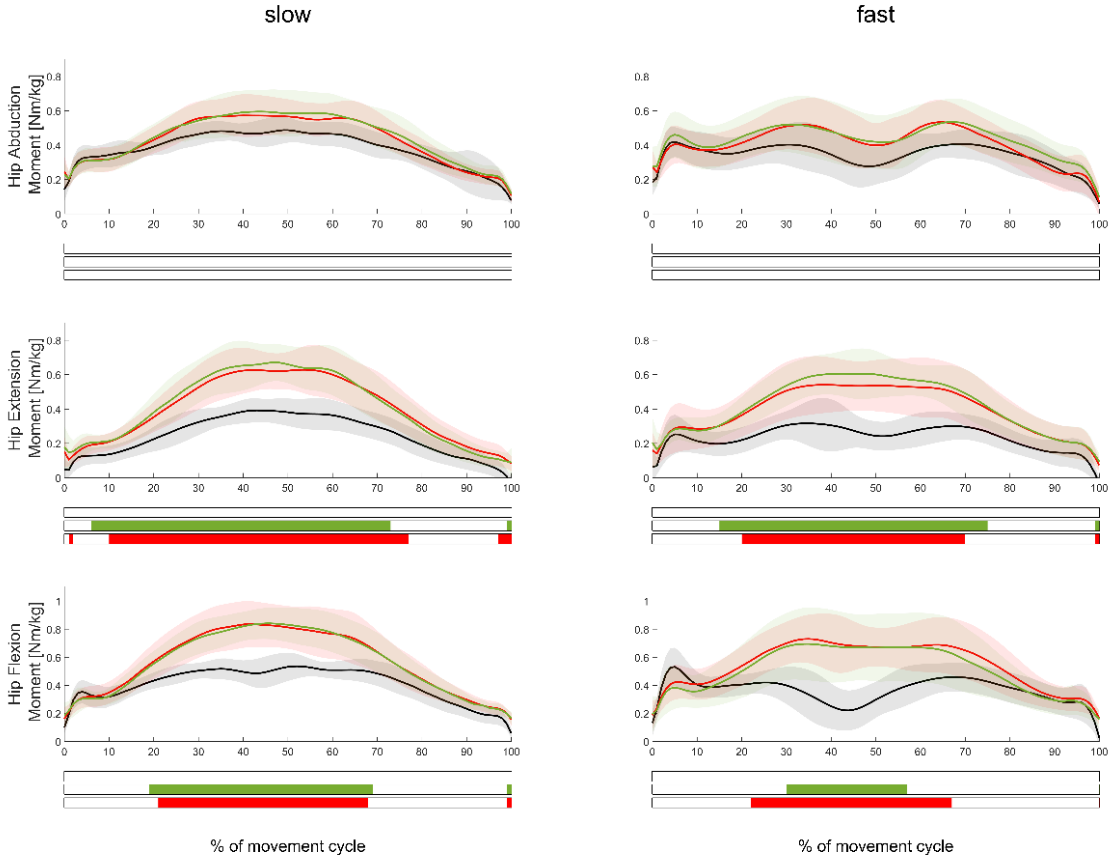

Figure S2. Mean (±SD) hip moment waveforms measured in the executing (movement) leg during hip abduction (top), extension (middle) and flexion (bottom) exercises, as well as during slow (left subplots) and fast (right subplots) velocity. Green, red and black waveforms represent the green (stiffer), red (softer) and no ERB, respectively. Colored bars beneath each plot indicate significant differences between waveforms, whereas the green, red and bue bars represent significant differences between the green versus no ERB, red versus no ERB and green versus red ERB, respectively.

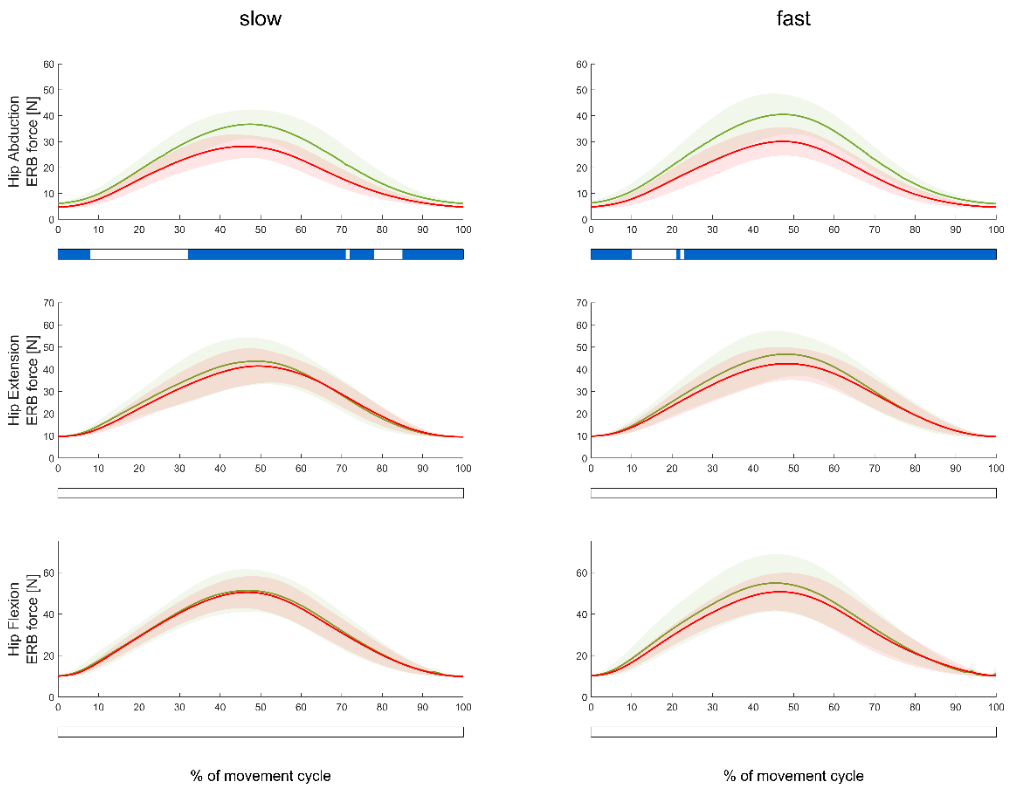

Figure S3. Mean (±SD) ERB forces during slow (left) and fast trials (right) measured in the red (red waveform) and green (red waveform) ERB during abduction (top), extension (middle) and flexion (bottom) exercises. Blue bars beneath each plot indicate significant differences between the forces of the red and green ERB.

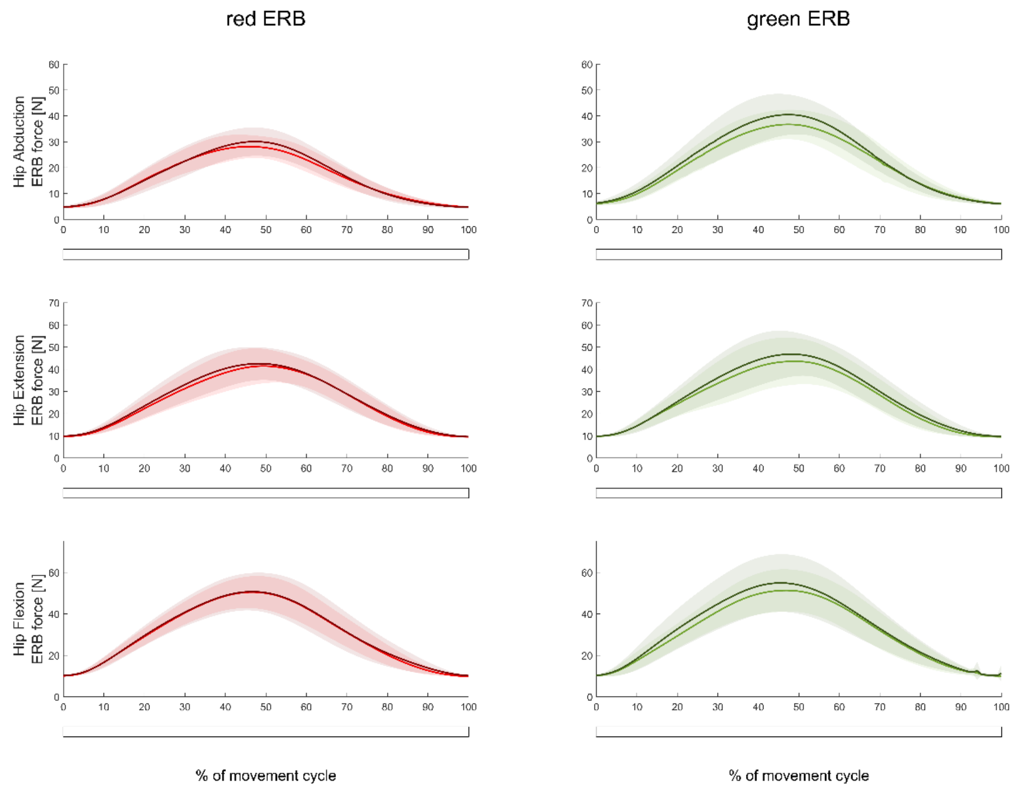

Figure S4. Mean (±SD) ERB forces during fast (dark red and dark green waveforms) and slow trials (light red and light green waveforms) measured in the red (left subplots) and green (right subplots) ERB during abduction (top), extension (middle) and flexion (bottom) exercises. ERB forces were not significantly different between the fast and slow movement executions.

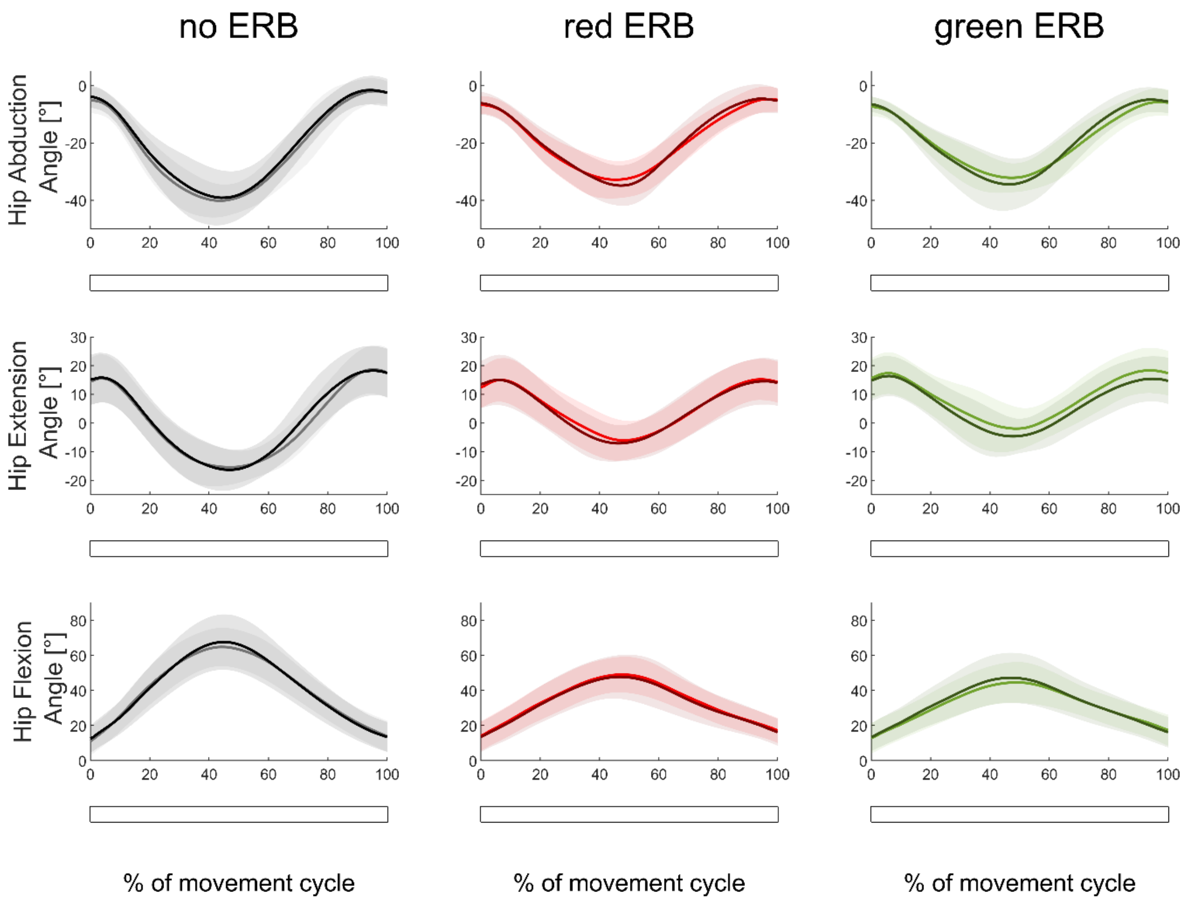

Figure S5. Mean (±SD) hip angle during fast (black, dark red and dark green waveforms) and slow trials (grey, light red and light green waveforms) measured in the trials without (left subplots), with red (middle subplots) and green (right subplots) ERB during abduction (top), extension (middle) and flexion (bottom) exercises. Kinematics were not significantly different between the fast and slow movement executions.

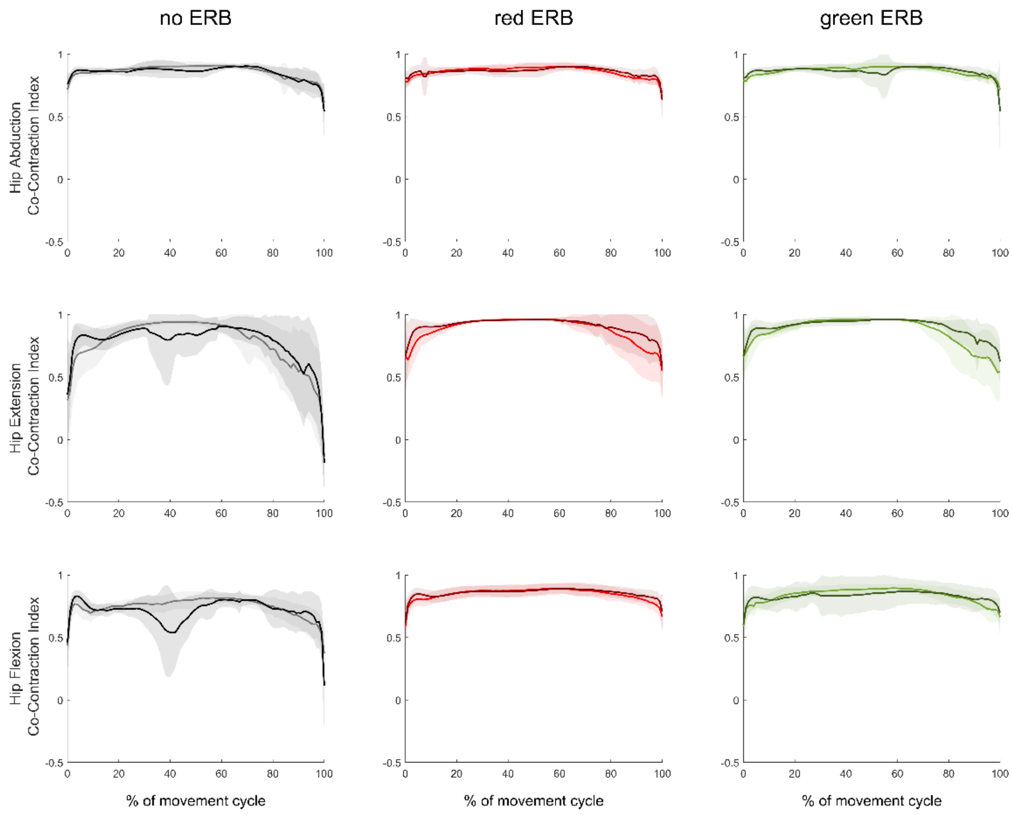

Figure S6. Co-contraction index calculated for the slow (bright waveforms) and fast (dark waveforms) movement.

3.14.3 Linear versus polynomial fitted curve to model the ERB force production

In our ERBs the relationship between force and elongation was not perfectly linear (Figure S7). Assuming a non-linear relationship and fitting a curve, i.e. 2nd degree polynomial curve, to our experimental data would have led to a better fit but this would not have affected our findings or conclusion (Figure S8 and S9). We chose a linear relationship to be consistent with previous publications (Hughes et al., 1999).

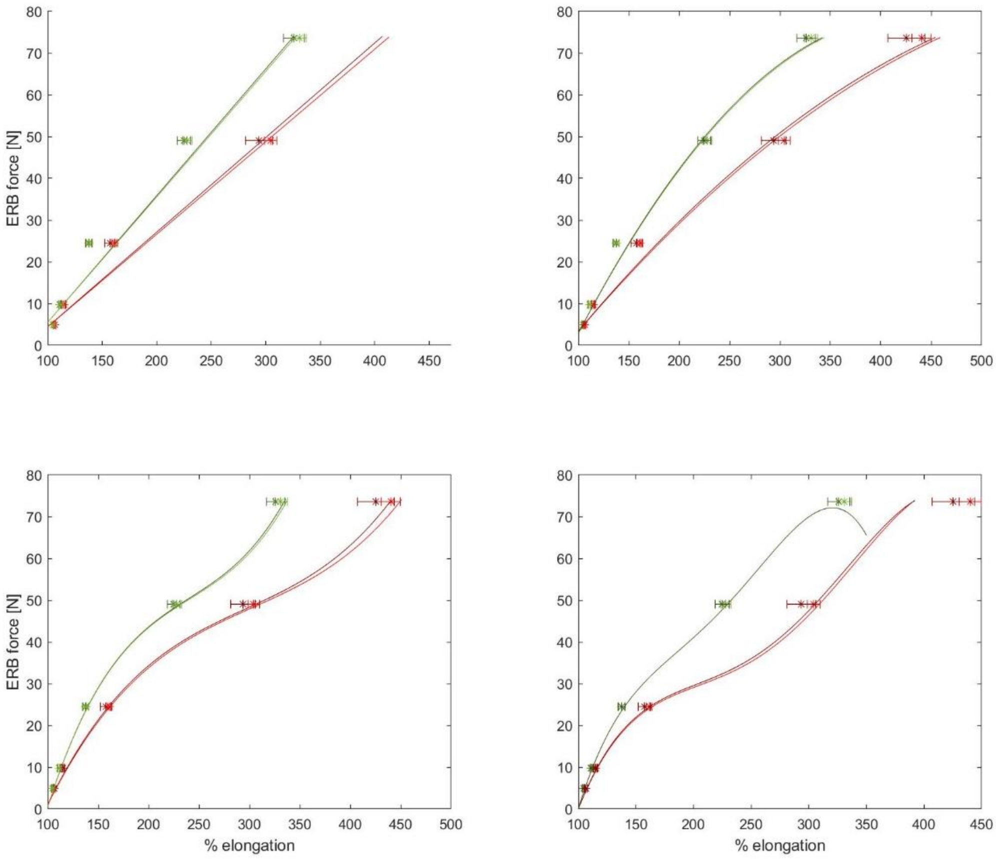

Figure S7. Mean force-elongation curves based on the assumption of a linear relationship (left top plot) and non-linear relationship. Right top plot: 2^{nd} degree polynomial curve, Button left plot: 3^{rd} degree polynomial curve, Button right plot: 4^{th} degree polynomial curve.

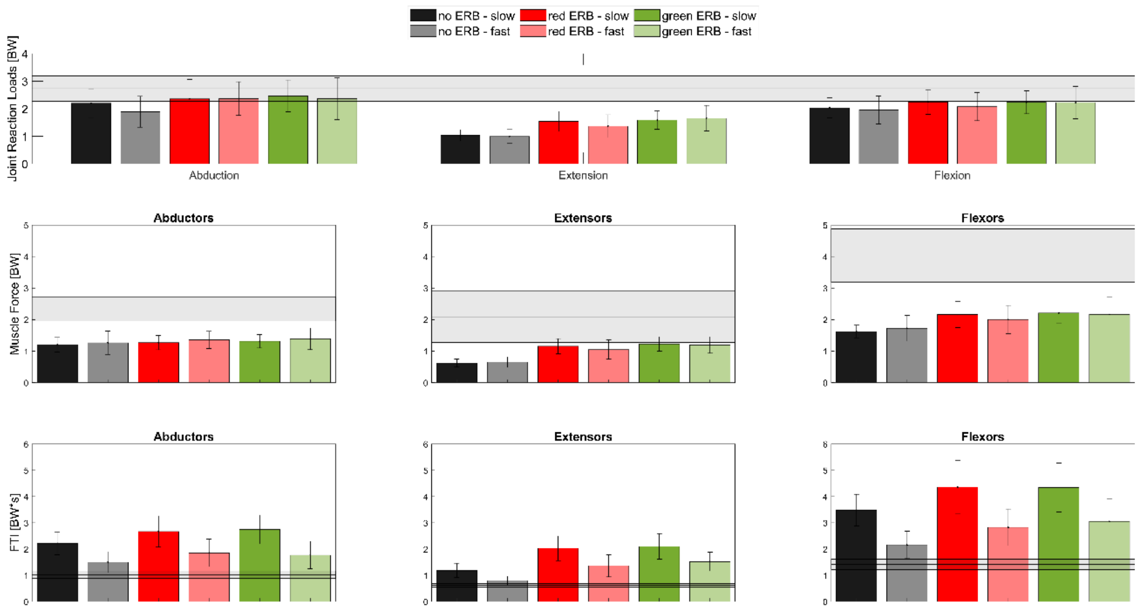

Figure S8. Results based on the assumption of a linear relationship between force and elongation of the ERB.

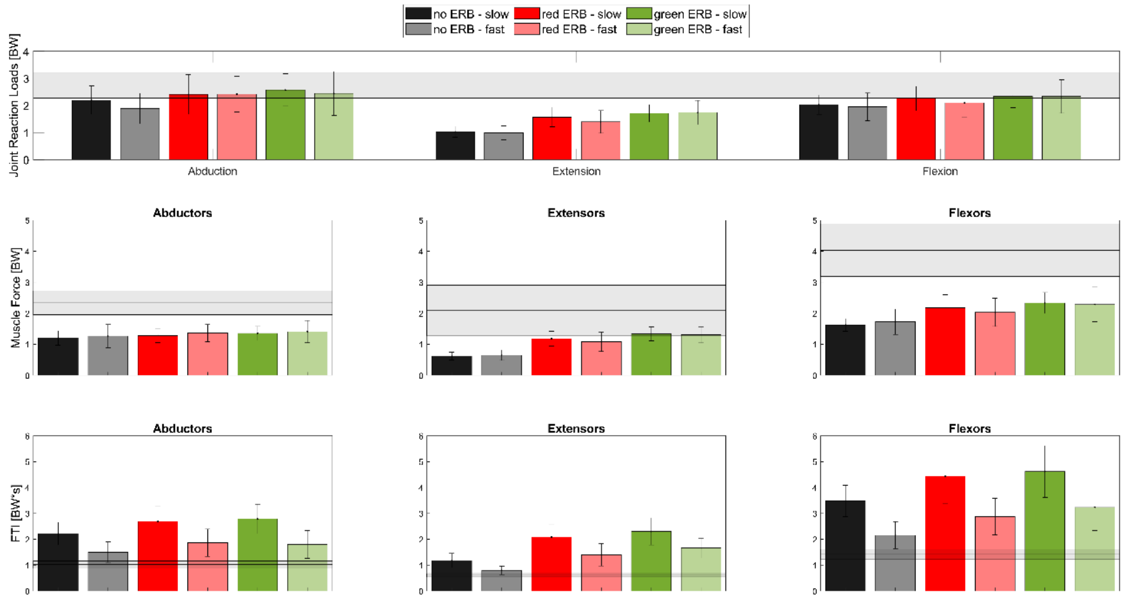

Figure S9. Results based on the assumption of a non-linear (2nd degree polynominal) relationship between force and elongation of the ERB.

Table S1. Peak HJCF and total muscle forces (FTI) during slow and fast execution velocity. The p-values are the result of the statistical comparison between slow and fast parameters. In all analyses the significance level was set to **α** = 0.05.

Peak HJCF								
		Slow			Fast			
		N	Mean [%BW]	SD [%BW]	N	Mean [%BW]	SD [%BW]	p-value
Abduction	Without	7	2.02	0.53	7	1.92	0.63	0.987
	Red	7	2.27	0.36	7	2.35	0.61	
	Green	7	2.39	0.54	7	2.41	0.59	
Flexion	Without	11	2.01	0.38	11	2.06	0.41	0.676
	Red	11	2.25	0.50	11	2.13	0.55	
	Green	11	2.21	0.44	11	2.19	0.64	
Extension	Without	10	1.06	0.21	10	0.95	0.24	0.878
	Red	10	1.52	0.38	10	1.46	0.26	
	Green	10	1.61	0.30	10	1.81	0.40	

Force-Time Integral (FTI)								
		Slow			Fast			
		N	Mean [BWs]	SD [BWs]	N	Mean [BWs]	SD [BWs]	p-value
Abduction	Without	7	2.27	0.50	7	1.58	0.37	<0.001
	Red	7	2.89	0.54	7	1.90	0.50	<0.001
	Green	7	2.94	0.51	7	1.96	0.40	<0.001
Flexion	Without	11	3.53	0.60	11	2.27	0.42	<0.001
	Red	11	4.30	1.08	11	2.88	0.73	<0.001
	Green	11	4.23	0.95	11	3.00	0.92	<0.001
Extension	Without	10	1.26	0.28	10	0.79	0.15	<0.001
	Red	10	2.05	0.41	10	1.48	0.23	<0.001
	Green	10	2.25	0.37	10	1.69	0.25	<0.001

Table S2. Peak muscle forces and force-time integral (FTI) of each respective muscle group as well as the peak HJCF measured during each exercise variant and during walking. The p-values are the result of the statistical comparison between the respective parameters measured during the exercise and walking trials. In all analyses the significance level was set to **α** = 0.05. n.s.=not significant different; WB=without band; R=red ERB; G=green ERB

			Peak Muscle Forces				**Force-Time Integral (FTI)**				**Peak HJCF**			
			N	**Mean [%BW]**	**SD [%BW]**	**p-val-ues**	**N**	**Mean [BWs]**	**SD [BWs]**	**p-val-ues**	**N**	**Mean [%BW]**	**SD [%BW]**	**p-value**
Gait	**Abd**		16	2.34	0.39	-	16	1.02	0.14	-				
	Flex		16	4.03	0.85	-	16	1.42	0.20	-	16	2.73	0.46	-
	Ext		16	2.09	0.82	-	16	0.61	0.07	-				
Abd	slow	**WB**	10	1.21	0.24	**<0.01**	10	2.22	0.46	**<0.001**	10	2.20	0.55	n.s.
	fast		8	1.25	0.42	**<0.01**	8	1.48	0.44	n.s.	8	1.85	0.62	n.s.
	slow	**R**	10	1.30	0.18	**<0.01**	10	2.82	0.46	**<0.001**	10	2.49	0.67	n.s.
	fast		8	1.33	0.36	**<0.01**	8	1.74	0.64	n.s.	8	2.19	0.72	n.s.
	slow	**G**	10	1.34	0.24	**<0.01**	10	2.88	0.44	**<0.001**	10	2.54	0.61	n.s.
	fast		8	1.33	0.44	**<0.01**	8	1.76	0.68	n.s.	8	2.18	0.84	n.s.
Flex	slow	**WB**	11	1.62	0.21	**<0.001**	11	3.53	0.60	**<0.001**	11	2.01	0.39	n.s.
	fast		12	1.84	0.30	**<0.001**	12	2.24	0.41	**<0.01**	12	2.07	0.39	**0.017**
	slow	**R**	11	2.13	0.44	**<0.001**	11	4.30	1.08	**<0.001**	11	2.25	0.50	n.s.
	fast		12	1.96	0.49	**<0.001**	12	2.79	0.76	**<0.01**	12	2.06	0.57	n.s.
	slow	**G**	11	2.15	0.30	**<0.001**	11	4.23	0.95	**<0.001**	11	2.21	0.44	n.s.
	fast		12	2.14	0.56	**<0.001**	12	3.01	0.88	**<0.01**	12	2.21	0.62	n.s.
Ext	slow	**WB**	12	0.62	0.13	**<0.05**	12	1.20	0.29	**<0.001**	12	1.04	0.20	**<0.001**
	fast		12	0.66	0.17	**<0.05**	12	0.79	0.15	**<0.05**	12	0.96	0.25	**<0.001**
	slow	**R**	12	1.13	0.22	**<0.05**	12	1.97	0.43	**<0.001**	12	1.52	0.35	**<0.001**
	fast		12	1.05	0.26	**<0.05**	12	1.37	0.36	**<0.05**	12	1.37	0.37	**<0.001**
	slow	**G**	12	1.23	0.24	**<0.05**	12	2.11	0.51	**<0.001**	12	1.58	0.34	**<0.001**
	fast		12	1.22	0.28	**<0.05**	12	1.57	0.36	**<0.05**	12	1.66	0.50	**<0.001**

3.14.4 Detailed results from the SPSS analyses

Second hypothesis:

Peak HJCF and the force-time integral (FTI) were compared between the slow and fast exercise executions.

Peak HJCF

Abd

$F(1, 6) = 0.00$, $p = 0.987$, partial eta squared = 0.00

No interaction band versus speed

Flex

$F(1, 10) = 1.86$, $p = 0.676$, partial eta squared = 0.018

No interaction band versus speed

Ext

$F(1, 9) = 0.025$, $p = 0.878$, partial eta squared = 0.003

Signif interaction band and speed $F(2, 18) = 6.173$, $p=0.009$, partial eta squared = 0.407

FTI

Abd

$F(1, 6) = 65.94$, $p < 0.001$, partial eta squared = 0.917

No interaction band versus speed

Flex

$F(1, 10) = 126.99$, $p < 0.001$, partial eta squared = 0.927

No interaction band versus speed

Ext

$F(1, 9) = 84.572$, $p < 0.001$, partial eta squared = 0.904

No interaction band versus speed

Post-hoc comparison $p < 0.001$ for all three comparisons

Third hypothesis:

Peak and total muscle forces but not peak HJCF of the executing leg will be higher compared to walking

Peak HJCF

Abd slow

$F(1.3, 11.7) = 1.644$, $p = 0.231$, partial eta squared = 0.154

Abd fast

$F(3, 21) = 4.396$, $p = 0.015$, partial eta squared = 0.386

Contrast $F(1, 7) = 5.287$, $p=0.055$ -> no significant post-hoc results

Flex slow

$F(3, 30) = 6.058$, $p = 0.002$, partial eta squared = 0.377

Contrast $F(1, 10) =8.666$, $p = 0.015$

Post hoc -> no signif difference

Flex fast

$F(3, 33) = 5.031$, $p = 0.006$, partial eta squared = 0.314

Contrast $F(1, 11) =10.53$, $p = 0.008$

Post hoc -> gait vs ohne fast $p=0.017$ (flex fast signif lower compared to walking)

Ext slow

$F(1.7, 18.3) = 57.892$, $p < 0.001$, partial eta squared = 0.84

Contrast $F(1, 11) =70.462$, $p < 0.001$

Post hoc -> gait vs all ext trials $p < 0.001$ (gait signif higher)

Ext fast

$F(3, 33) = 56.744$, $p < 0.001$, partial eta squared = 0.838

Contrast $F(1, 11) =127.339$, $p < 0.001$

Post hoc -> gait vs all ext trials $p < 0.001$ (gait signif higher)

Peak muscle force

Abd slow

$F(1.3, 11.7) = 34.919$, $p <0.001$, partial eta squared = 0.795

Contrast $F(1, 9) = 40.999$, $p<0,001$

Post hoc -> gait vs all abd trials $p < 0.01$ (gait signif higher)

Abd fast

$F(3, 21) = 24.949$, $p <0.001$, partial eta squared = 0.781

Contrast $F(1, 7) = 46.104$, $p<0,001$

Post hoc -> gait vs all abd trials $p < 0.01$ (gait signif higher)

Flex slow

$F(1.29, 12.923) = 49.285$, $p < 0.001$, partial eta squared = 0.831

Contrast $F(1, 10) =54.129$, $p <0.001$

Post hoc -> gait vs all flex trials $p < 0.001$ (gait signif higher)

Flex fast

$F(3, 33) = 35.226$, $p < 0.001$, partial eta squared = 0.762

Contrast $F(1, 11) =51.701$, $p < 0.001$

Post hoc -> gait vs all flex trials $p < 0.001$ (gait signif higher)

Ext slow

$F(1.12, 12.32) = 29.325$, $p < 0.001$, partial eta squared = 0.727

Contrast $F(1, 11) =25.675$, $p < 0.001$

Post hoc -> gait vs all ext trials $p < 0.05$ (gait signif higher)

No band $p < 0.001$; red $p = 0.005$; green $p = 0.013$

Ext fast

$F(1.29, 14.22) = 24.575$, $p < 0.001$, partial eta squared = 0.691

Contrast $F(1, 11) =24.578$, $p < 0.001$, partial eta squared = 0.691

Post hoc -> gait vs all ext trials $p < 0.05$ (gait signif higher)

No band $p < 0.001$; red $p = 0.007$; green $p = 0.010$

FTI

Abd slow

$F(3, 27) = 73.480$, $p <0.001$, partial eta squared = 0.891

Contrast $F(1, 9) = 111.854$, $p<0{,}001$, partial eta squared = 0.926

Post hoc -> gait vs all abd trials $p < 0.001$ (gait signif lower)

Abd fast

$F(3, 21) = 7.071$, $p = 0.002$, partial eta squared = 0.503

Contrast $F(1, 7) = 11.447$, $p=0.012$, partial eta squared = 0.621

Post hoc -> no signif. difference

Flex slow

$F(3, 30) = 51.567$, $p < 0.001$, partial eta squared = 0.838

Contrast $F(1, 10) =95.359$, $p <0.001$, partial eta squared = 0.905

Post hoc -> gait vs all flex trials $p < 0.001$ (gait signif higher)

Flex fast

$F(2.074, 22.817) = 21.138$, $p < 0.001$, partial eta squared = 0.658

Contrast $F(1, 11) =35.490$, $p < 0.001$, partial eta squared = 0.763

Post hoc -> gait vs all flex trials $p < 0.01$ (gait signif lower)

Ext slow

$F(3, 33) = 89.323$, $p < 0.001$, partial eta squared = 0.890

Contrast $F(1, 11) =124.815$, $p < 0.001$, partial eta squared = 0.919

Post hoc -> gait vs all ext trials $p < 0.001$ (gait signif lower)

Ext fast

$F(3, 33) = 46.458$, $p < 0.001$, partial eta squared = 0.809

Contrast $F(1, 11) =76.351$, $p < 0.001$, partial eta squared = 0.874

Post hoc -> gait vs all ext trials $p < 0.05$ (gait signif lower)

No band $p = 0.047$; red $p < 0.001$; green $p < 0.001$

3.15 Acknowledgments

The authors would like to acknowledge Liam Strasser and David Deimel for their help during data collection and to all participants who took the time to participate in this study. Furthermore, the authors would like to thank Ass-Prof Dr. Peter Gröpel for his support during their statistical analyses.

4 Discussion

As mentioned in the paper above, the overall objective of the conducted research was to further our understanding of the loading behaviour of the hip during execution of rehabilitative exercises for hip OA. In order to do so load characterizing parameters such as HJCF and muscle forces were measured during the exercises, analysed through use of MSK simulations and, in order to put them in relation, compared with those found during walking. The motivation for the research was to enable practitioners working in the fields of rehabilitation and prevention to use the resulting information to make informed and effective recommendations for rehabilitation exercises to treat OA in the hip, whilst minimising the associated risks, such as overloading the joint and thus contributing to OA progression.

Three hypotheses were made at the beginning of the research, all three of which have been adequately addressed in the paper. In light of the first hypothesis: muscle forces and JCF are higher when using a stiffer (green) resistance band compared to a softer (red) resistance band and no resistance band. Our findings showed that regardless of resistance level, i.e. whether using a stiffer or a softer ERB, muscle forces measured were higher in exercises in which ERBs were used than those resulting from exercises that were conducted using the participants own body weight. This is not surprising as with increasing training load, which in the case of study 1 would be the ERBs, it stands to reason that the muscle forces required to perform the movement would also increase. As the movements were specifically selected to stress certain hip muscles, it follows that the targeted muscles would also need to exert higher muscular forces at higher loads. However, in contrast to this conclusion, the results of the comparison between the softer and stiffer ERB showed that while a significant difference can be found between loaded and unloaded exercise performance in terms of HJCF and target muscle forces, the same cannot be said in relation to the different types of ERB used in the study, which partially contradicts our first hypothesis. In light of the results our first hypothesis cannot be fully confirmed nor discarded.

Our the second hypothesis: movement execution with a higher velocity will increase peak hip JCF but decrease total muscle forces, could only be party confirmed. While the total muscle forces, measured in the study as FTI, was shown to be consistently lower during the exercises performed with the faster velocity versus those executed with the slower velocity. A possible explanation for this could be that due to the slower execution stabilizing muscles of the hip have to be more active for a longer period of time than compared with a faster execution, thus increasing the total muscle work required to perform the movement. However, the results of the study did show that while FTI did significantly change with the variance in velocity, the HJCF did not, thus partly contradicting our second hypothesis.

Likewise, the third hypothesis, that peak and total muscle forces but not peak hip JCF of the executing leg will be higher compared to walking, could not fully be confirmed. While the data showed that the FTI was measured to be significantly higher during exercises using ERBs as a loading modality compared to those found during the recorded gait trials, the same could not be said with regard to peak muscle forces as well as peak HJCF. The paper showed that both peak muscle forces and peak HJCFs were either similar or significantly lower during the recorded gait trials in comparison to the exercise trials in which an ERB was used.

Although the hypothesis could not be fully confirmed, the comparison between the parameters found during the exercises - whether with or without loading modality - and the gait trials can show a relationship between the two conditions and thus provide a better understanding of the loading behaviour of the joints and target muscles. Especially considering the body of research done on gait and how well understood walking is as a metric, the comparison ensures that the exercises can be better rated in terms of their stress. Considering the results of the study, i.e. that the HJCF measured during the exercise trials were surprisingly low and did not exceed the values found during walking, it can be concluded that if walking can be recommended as part of a rehabilitation of hip OA or is considered a safe form of exercise, the exercises presented, loaded or unloaded, can be performed without concern that they may worsen the progression of the disease. This could provide practitioners in the field of OA rehabilitation further confidence while recommending similar exercises as part of a rehabilitative exercise regime.

Based on the conclusions from the study regarding preferable exercises execution velocity, it seems that variations in movement velocity during exercise performance is acceptable and does not change resulting HJCF. This would suggest that, when merely considering joint loading, velocity can be adapted to patient preference and does not need to be too stringently controlled. That being said, a slower execution did show a significantly higher FTI, meaning that slower execution speeds could be more expedient as it could provide more overall muscle work and thus could facilitate more muscle growth. Furthermore, if overall muscle work is controlled, it would be preferable to the higher execution speeds as a slower speed would mean a higher FTI in less repetitions of the exercises and thus less loading cycles and less resulting overall joint stress. This conclusion could help avoid unnecessary exacerbation of the diseased joint.

Interestingly, the study did not find a significant difference in the measured parameters with respect to the different ERB types, i.e. the red, soft band and the green, stiffer band. This lack of contrast could be due to the difference in generated resistance between the two ERBs being too small. It is possible that if there was a greater difference in load behaviour, the measured parameters would show greater differences. The problem with this is that using a larger variation in ERBs would not represent the reality of clinical practice as accurately. As mentioned in the sections above, the bands were specifically selected as they are likely to be used in rehabilitation settings. Stronger, more resilient bands would most likely be too stiff for OA patients, especially considering the average age of those being treated for OA. In addition, softer bands might not provide sufficient resistance to see a difference between bodyweight exercises and those loaded with ERBs. However, due to the lack of difference, the study shows that the type of ERB is not as important as initially thought, implying that considering the elastic nature, the choice of ERB is better adapted to the patient-specific ROM, as the magnitude of the displacement amplitude is far more impactful regarding acting load.

During the experimental procedure, continuous measurement of the ERBs before and after each subject ensured that the stress-strain behaviour of the ERBs did not change significantly over all measurements. However, despite careful handling of the ERBs used, the green ERB tore during the experimental session of the 13th subject. As a result, a new initial measurement of the stress-strain behaviour of the new ERB was made. It was found that there was no significant difference between the torn ERB and its replacement. While it is unlikely, this could have had an influence on preceding measurements. The cause of the tear was attributed to the adhesive of the retroreflective surface markers that were placed on the ERBs to measure their length displacement during the trials. The adhesive caused the material of the ERB to become brittle and the repeated stress caused by the stretching caused the ERB to tear. Further research using similar materials and methods could use the information in order to avoid similar incidents.

One of the biggest limitations of the chosen experimental design is the different fixation variants of the ERB to the subjects. For practical reasons, which are mentioned above, the fixation variant of the flexion and extension trials differs from that of the abduction trials. As a result of this difference, the lever arm of the applied ERB force in relation to the hip changes by the length of the subject's lower leg. This means that the load is different for the same amplitude and makes it unsuitable for comparison. However, it should be noted that the ROM of the patients was not fixed, and the ERB loading depends on the ROM of the patients, so it is likely that if the fixation variants were the same, the patients would still deflect up to the same maximum ERB force. That being said, for further analysis of the differences this assumption would need to be tested in more detail in further research.

It could well be argued that the biggest limitation of the study presented is the subject group. The aim of the study was to gather information relevant to practice in order to provide experts with a better scientific basis for rehabilitation recommendations. The problem is that the subjects of the studies do not correspond to the target group of the rehabilitation recommendations. The subjects of the study were all young, healthy individuals without any medical conditions. If the study had been carried out on older subjects with hip OA, the results might have differed. Such a study, specifically with the target group of people suffering from hip OA, would be necessary to validate the results obtained in this study.

With regard to the validation of the HJCF the study goes on to show that mean HJCF waveforms of exercises using no ERB for external loading were comparable with those measured using in vivo instrumented hip implant. This was true for all participants in both movement and supporting leg. Similarly to the findings presented on OthroLoad the HJCF measured during the exercises without external loading were higher in the supporting leg when compared with the HJCF recorded in the movement leg. As part of the qualitative comparison waveform shape as well as maximum HJCF values were compared. This comparison shows that, with exception of the HJCF in the movement leg during hip abduction, both shape and maximum values were comparable with each other. During the visual comparison, apart from minor differences between HJCF waveforms in hip abduction, the values found on OrthoLoad displayed a reasonable level of conformity with those measured during the experimental trials. Although visual comparison of the values with those of OthroLoad values is not as accurate as a statistical test for significant differences, they served as a reference and provided more confidence in the correctness of the simulation values. However, it is worth mentioning that one limitation of the Orthoload values is that the sample size of the experiments conducted by OthroLoad is only one participant. The origin of these differences could well be explained by taking into account the variations in hip kinematics and execution speed depending on the subjects during the movements. Furthermore, the differences caused by the different data collection methods, in this case in vivo measurements and simulated results, must not be disregarded.

5 Conclusion

The aim of the present work was to quantify the muscle forces of the exercise-specific loaded muscles as well as the joint loads during typical rehabilitation exercises in hip OA performed with ERBs. We hoped to answer the question to what extent the HJCF measured with MSK simulations during the exercises differ from each other and from those of a normal gait cycle. In this respect, several conclusions can be drawn based on the information from our results. Firstly, the study showed that the type of ERB used, whether it be stiff or soft, had a minor impact on the peak muscle forces and HJCF during exercise execution. Secondly, similar to ERB type, execution velocity had a minor impact on peak muscle forces and HJCF, however unlike ERB type, execution velocity does affect the total muscle force required for an exercise. Thirdly, when compared with parameters measured during walking the exercises that were performed without an ERB showed similar or lower peak HJCF as well as lower muscle forces. The fourth conclusion being that adding ERB to the performed exercises increased the peak muscle forces and HJCF but even then, the estimated values did not exceed those of a normal gait cycle. The fifth and final conclusion being that the FTI did exceed the values of the same parameter estimated during walking.

Considering the research question and the aim of the project, the study fulfilled its purpose and demonstrated the effects of rehabilitative hip exercises on hip joint loading and surrounding muscle forces. The results of the work can contribute to the growing body of rehabilitation research related to the rehabilitation and management of OA. It is hoped that as knowledge of such exercises increases, clinicians working in this field will be able to base their interventions and recommendations on this type of information to better prevent rapid progression of OA.

5.1 Implications and Outlook

While the results of the study contribute to rehabilitation research, further questions arise in this area. With regard to the data collected but not used in the paper, e.g. the motion capture and force plate data from the squat trials as well as from all trials conducted using the left leg as the movement leg instead of the standing or balancing leg, it was not processed as part of the project as it was also deemed to surpass our scope. This data could present an opportunity to further delve into more detail regarding the relationship and correlation between HJCF, muscle force and rehabilitation exercises. Furthermore, the question as to what the acceptable upper limit of HJCF is experienced by hip OA patients during exercises still remains unanswered. It would also make sense to carry out similar studies to the one presented, but with an adapted, more practice-oriented group of participants. As mentioned in the limitations, a different result would be possible if the subjects were more similar to the age group of OA sufferers. Similarly, other results would be conceivable if the number of subjects was larger. Long-term effects of rehabilitation exercises are particularly worthy of research. A long-term study that examines HJCF, muscle strength and OA progression in more detail would give us an insight into how such exercises can curb the progressive development of such diseases and could lead to even more effective and appropriate design of interventions.

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ASIS anterior superior iliac spine

BPM Beats per minute

BW Body weight

DoF Degrees of freedom

EMG Electromyography

ERB Elastic resistance band

FTI Force-time integral

HHD hand-held-dynamometer

HJCF Hip joint contact force

IK inverse kinematics

ID inverse dynamics

JCF Joint contact force

MSK Musculoskeletal Simulation

NCD non-communicable disease

NSAIDS non-steroidal anti-inflammatory drugs

OA Osteoarthritis

PC performance criterion

QoL quality of life

RMS Root mean square

ROM Range of motion

RPE Rate of perceived exertion

SD Standard deviation

SO Static optimisation

SPM Statistical Parametric Mapping

TEP total endoprosthesis

VSK labelling skeleton

VST labelling skeleton template

YLD Years lived with disability