Muscle forces in the lower extremities during strength training for strength enhancement, prevention and rehabilitation of an anterior cruciate ligament rupture

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Muscle forces in the lower extremities during strength training for strength enhancement, prevention and rehabilitation of an anterior cruciate ligament rupture
Muscle forces in the lower extremities during strength training for strength enhancement, prevention and rehabilitation of an anterior cruciate ligament rupture

A thesis submitted to attain the degree of

DOCTOR OF SCIENCES of ETH ZURICH

(Dr. sc. ETH Zurich)

presented by

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2016
Statement of Originality

I hereby confirm that I am the sole author of the written work here enclosed and that I have compiled it in my own words. Corrections of form and content were partially written with the support of the supervisors.

Florian Schellenberg
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From the very beginning of my academic career at the institute for biomechanics, Dr. Dr. Silvio Lorenzetti supported me and fortunately gave me the opportunity to successfully complete my doctoral study. He is not only an excellent scientific mentor but also a unique personality and has become a business partner and real personal friend. His advice on both scientific inputs and personal enhancement has been priceless. Silvio, I would like to take this opportunity to deeply thank you for your advice, your ideas, and your social dedication as a group leader but also for your extraordinary cooperation during the whole time here at the institute.

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Abstract

Strength training is a training method focussed primarily on strengthening a specific muscle and to increase its size. In many sports, strength training is a key requirement to enhance performance, but strength training is also commonly performed to improve and maintain general health, or for aesthetic reasons. Additionally, strength training can be applied to prevent various injuries or to recover from an already sustained injury. In all cases, the employed strength training needs to be performed in a very specific and controlled manner to achieve the desired effects in a safe, fast, and efficient way.

The adaptability of a muscle is mainly determined by the loading conditions acting on the muscle itself. High loading conditions lead to muscle strengthening and synthesis but too high loading conditions can cause damage or inflammation. Consequently, knowledge of the internal loading conditions during strength exercises is a key requirement to avoid injuries and to ensure positive tissue adaptation. This knowledge of internal loading conditions would also allow coaches or therapist to develop efficient, individualized training programs that improve performance, health, and mobility, while minimizing the risk of injury at the same time. Unfortunately, these internal loading conditions are hardly measurable directly and are thus unknown for most strength exercises. This work aimed to address the question of how to determine loading conditions, especially the individual muscle forces, and corresponding joint angles during various strength exercises. Based on these findings, recommendations for training and rehabilitation programs have been developed.

A systematic literature review investigated current methods to calculate and determine muscle loading during strength exercises. While a direct measurement of the in vivo muscle forces remains infeasible with current techniques, different approaches to estimate muscle forces indirectly were identified and classified into three method groups: forward dynamics simulations, inverse dynamics techniques, and alternative methods such as EMG-driven modelling. While forward
dynamic simulations lead to considerable new insights into muscular coordination, strength, and power during dynamic ballistic movements, inverse dynamic techniques and EMG-driven modelling allow an accurate estimation of the loading conditions during low-speed movements such as strength exercises.

Since inverse dynamic techniques allow estimating loading conditions during strength exercises, a quasi-static inverse dynamic approach was used to calculate the external net moments in the knee, hip, and lower back joints during various strength exercises and execution forms. Specifically, this work investigated the following strength exercises: deadlifts, goodmornings, split squats, and back hyper extensions. The computed external moments and corresponding joint angles for the investigated strength exercises were summarised and discussed in their entirety. This summary allows trainers and therapist to recommend an appropriate strength exercise dependent on the individual aims of the athletes and patients. The evidenced-based recommendation leads to safe trainings and rehabilitation process and adapts musculature structured as desired.

The calculation of external net moments is useful to estimate the loads acting on the joint during a strength exercise, but a more precise analysis of individual musculature during a strength exercise is required, since loading conditions in individual muscles do not necessarily match with external moments. Strength training aims to strengthen individual muscles, and hence loading conditions of each muscle need to be known. Since the number of muscles crossing a joint is greater than the number of degree of freedom of the joint, the system is mathematically over-determined. This over-determination requires optimization techniques to estimate individual muscles forces. Musculoskeletal simulations offer the opportunity to estimate internal loading conditions such as muscle forces through the application of optimization criteria. For accurate muscle force estimation, adequate recommendations on the technique of the simulation as well as validated musculoskeletal simulations during strength exercises are crucial. This thesis addresses both aspects through investigating the robustness of different simulation techniques, followed by a validation of subject-specific standard OpenSim reference models by means of instrumented arthroplasties measuring internal joint reaction forces during squatting.

Based on the findings of the robustness study, recommendations to achieve a robust musculoskeletal simulation are as follows: 1) include functionally determined centres of rotations, 2) include skin makers based on an automated weighting procedure that minimizes soft tissue artefacts, and 3) include pre-calculated joint angles into the simulation.

The validity of the standard reference model was assessed in a study on 6 subjects with in-
Abstract

Instrumented total knee arthroplasty. The subjects were asked to perform squats while the tibiofemoral joint contact forces were measured. Accuracy between measured and calculated joint contact forces resulted in an average peak error of approximately 60%. The error linearly correlates with corresponding knee flexion angle, resulting in an underestimation of the calculated joint contact forces during an upright posture and overestimation during a posture including deep knee flexion angle. A linear correlation of the error could be shown. Further investigation is required to identify the origin of the analysed error, to enhance the accuracy of the model and thus to reduce the error. To reduce the error, it is recommended to focus on accurate representations of muscle paths and corresponding moment arms for further developments of the model, especially during extreme joint angles.

Based on the results of the robustness and validation studies, an adequate technique to simulate musculoskeletal models was developed, and the capabilities and limitations of such simulations were identified. Using the developed technique, measured kinematic and kinetic data of the exercises deadlift, goodmornings and split squats were used to estimate individual muscle forces in the lower extremities acting during the exercises. Results indicate that individual muscle forces are not necessarily in agreement with external moments, and that these muscle forces are strongly dependent on the specific exercise as well as its executions. Using estimated muscle forces combined with well-known injury risk factors, recommendations to prevent anterior cruciate ligament (ACL) ruptures were identified. To prevent an ACL tear, the hamstrings to quadriceps ratio needs to be high. The estimated muscle forces indicate that goodmornings are favourable over deadlifts and split squats to enhance the aforementioned ratio and thus prevent ACL injuries. Furthermore, estimated forces of individual muscles during different strength exercises now allow athletes and patients to individually design their training to achieve a desired adaptation of soft tissues according to the aim of the training.

The present work investigated in methods of analysis to determine loading conditions acting during different strength exercises. External net moments as well as corresponding joint angles of the exercises deadlifts, goodmornings, split squats, and back-hyper-extension were calculated using an inverse dynamic approach. Individual muscle forces arising during the three former mentioned exercises were estimated using an optimization technique of a musculoskeletal simulation. Accuracy and limitations of said musculoskeletal simulation were analysed and discussed and require further development of the examined standard model. Using the data and conclusions presented in this work, trainer and therapists are able to design evidence-based, targeted, and safe training, prevention and rehabilitation programs.
Zusammenfassung

Krafttraining ist eine weit verbreitete Sportart um primär die Muskelkraft und den Muskelquerschnitt zu vergrößern. In vielen Sportarten ist Krafttraining eine wesentliche Voraussetzung, um die sportliche Leistung zu steigern, aber Krafttraining fördert auch allgemein gesundheitliche Aspekte und steigert die Ästhetik. Zusätzlich zur Leistungsförderung im Sport wird Krafttraining zur Verletzungsprävention angewendet, aber auch in einem Rehabilitationsprogramm bei einer bestehenden Verletzung eingesetzt. In jedem Fall muss das Krafttraining spezifisch und kontrolliert ausgeführt werden, damit die gewünschten Effekte sicher, schnell und effizient erreicht werden.


In einer systematischen Literaturarbeit wurden aktuelle Methoden zur Berechnung und Bestimmung von internen Belastungszuständen während Kraftübungen geprüft. Mit der aktuellen Technik ist es bis heute nicht möglich, die Muskelkräfte direkt in vivo zu messen, aber


Die Resultate der Robustheitsstudie sowie der Validierung wurden einerseits gebraucht, um eine angemessene muskuloskeletale Simulationstechnik zu entwickeln und andererseits, um die Mög- lichkeiten und Limitationen abschätzen zu können. Die entwickelte Simulationstechnik wurde auf die gemessen kinematischen und kinetischen Daten der Übungen Kreuzheben, Good Mornings und Ausfallschritte angewendet, um Kräfte der einzelnen Muskulaturen der unteren Extremitä-
ten während diesen Kraftübungen abzuschätzen. Es konnte gezeigt werden, dass die spezifischen Muskel-Kräfte nicht zwingend mit den berechneten externen Momenten übereinstimmen. Die einzelnen Muskelkräfte variieren sowohl stark zwischen den verschiedenen Übungen als auch zwischen verschiedenen Ausführungsformen. Die Kombination dieser Resultate mit bekannten, verletzungsbedingten Risikofaktoren ermöglichte die Identifikation von Empfehlungen bezüglich Präventionsmassnahmen für vordere Kreuzbandrupturen. In der Prävention und Rehabilitation von vorderen Kreuzbandrupturen sollte das Verhältnis der hinteren zur vorderen Oberschenkel-
muskulatur erhöht werden. Anhand der analysierten Resultate ist in diesem Zusammenhang die Ausführung der Good Mornings vor dem Kreuzheben und den Ausfallschritten als Präventions-
masse zu favorisieren. Des Weiteren können Mithilfe der berechneten Muskelkräfte und den entsprechenden Gelenkwinkel nun individuell geeignete, dem Individuum angepasste Kraftübun-
gen identifiziert und ausgeführt werden. Dies führt zu einer trainingsrelevanten, zielgerichteten
Adaption der Weichteile.

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<tr>
<td>ACL</td>
<td>Anterior cruciate ligament</td>
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<td>BE</td>
<td>Back extension</td>
</tr>
<tr>
<td>BifemLh</td>
<td>Musculus biceps femoris long head</td>
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<tr>
<td>BifemSh</td>
<td>Musculus biceps femoris short head</td>
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<tr>
<td>BW</td>
<td>Body weight</td>
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<td>CAD</td>
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<td>Centre of rotation</td>
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<td>Deadlift</td>
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<td>DoF</td>
<td>Degrees of freedom</td>
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<td>Electromyography</td>
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<td>fCoR</td>
<td>Functional centres of rotation</td>
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<td>Ground reaction forces</td>
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<td>Hamstrings to quadriceps ratio</td>
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<td>Joint contact forces</td>
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<td>MS</td>
<td>Musculoskeletal</td>
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<td>MVIC</td>
<td>Maximal voluntary isometric contraction</td>
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<tr>
<td>OMC</td>
<td>Optical motion capture</td>
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<td>PCSA</td>
<td>Physiological cross sectional area</td>
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<td>Parallel-elastic element</td>
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<tr>
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<td>Split squat</td>
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<td>TKA</td>
<td>Total knee arthroplasty</td>
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<tr>
<td>VasInt</td>
<td>Musculus vastus intermedius</td>
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<td>VasMed</td>
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Chapter 1

Introduction

Strength training is common in sport exercise to develop the strength and the size of skeletal muscular structures. In many sports, strength training is an important factor to enhance performance, but it is also performed to improve and maintain general health. Due to the enhanced performance and healthy aspects, strength training has become widely popular. An estimated 4.9% of the Swiss population performs strength training at least twice a week, which corresponds to an increase in 1.8% from 2008 [1]. Similar results were found in the US, where an increasing number of young high school athletes participate in strength training (approximately 16000 in 2004/2005). Apart from the enhancement of sport performance, strength training can also be performed to prevent injuries, help recover from already sustained injuries, and for aesthetical reasons. On the contrary, physical inactivity can lead to negative health effects, such as obesity, loss of muscle tissue, osteoporosis, and others. These negative health effects do not only place a considerable burden on the afflicted individual but also on the public health sector. For example, it is estimated that the public health costs due to the effects of physical inactivity are about £8.2 billion a year in the UK alone [2]. To avoid negative health effects, the World Health Organisation (WHO) recommends in their “global strategy on diet, physical activity, and health” to strengthen major muscle groups two or more days a week [3].

Biological structures - including musculature - are able to adapt during the whole lifetime of a human to required demands. Strength training uses this adaptability to strengthen a muscle but also to enhance anaerobic endurance and size of musculature in a fast, safe and efficient manner. To achieve such a fast, safe, and focussed adaption, the correct exercises have to be chosen, tailored to the individual and a specific training goal. Loading conditions such as acting
forces and moments but also joint angles are key factors regarding the adaptation of a tissue and vary from exercises to exercise. Here, especially internal loading conditions, e.g. muscle forces or ligament loads, are important and should therefore be taken into consideration when choosing the exercise. There is a variety of exercises and execution possibilities to train a specific muscle or muscle group, each leading to different loading conditions and movement patterns. As mentioned, such loading conditions during strength training are very important to achieve efficient tissue adaption [4–7] but can also lead to injury or inflammation when inappropriate loading conditions are applied. Therefore an optimal balance in loading of the tissues is essential.

However, individualities of the athletes and patients such as strength and deficits as well as acting forces and kinematic patterns during different movement tasks, especially in strength exercises, are currently not well understood and researched. Therefore, better knowledge of the internal loading conditions during daily activities and strength exercises is a key requirement to develop more efficient individualized training programs to improve performance, health, and mobility, while minimizing the risk of injury at the same time. Because in vivo measurements of internal loading conditions, such as muscle forces, are highly limited with current techniques, musculoskeletal simulations are developed and widely accepted to calculate these internal loading conditions using an optimization criterion. Unfortunately, loading conditions, such as muscle forces and joint moments, remain unknown for most strength exercises and their variations of execution.

### 1.1 Application of Strength Training

Strength training is often used to enhance sport performance of athletes, in rehabilitation processes to recover from injuries, but also to prevent future injuries. Strength training targets soft tissue and musculoskeletal structures and is associated with a number of positive aspects, such as prevention of muscle atrophy [8], improvement of the general health [9], enhancement of bone mineral density, lean body mass with a reduction of body fat [10], and others. On top of these positive aspects, strength training is associated with a low risk of injury when compared to other sport activities [11, 12], making up only 0.7% of all sports related injuries [13]. When an injury related to strength training occurs, the prime causes are usually overloading (46%) and wrong execution of the exercise (21%). The most affected injured parts of the body are shoulder (24%), back (17%), thigh (11%), and knee (9%) [14]. Overloading seems to be the prime factor to cause injuries during strength training, but high loading is required to achieve
targeted adaptation such as increased strength in desired muscles. Therefore, a key criterion for a successful strength exercise is that the loading of the trained joint and muscles is high enough over specified ranges of motion (RoMs) to achieve a positive adaptation process, while staying low enough to avoid injuries. Additionally, the loading in all other joints should stay as low as possible to avoid injuries in these joints and to ensure training of the target muscles only.

While the kind of exercise and its execution are clearly the main determining factors of the resulting joint loadings [15], quantitative information on internal loading conditions for specific exercises remain hard to obtain [16, 17]. Because the biomechanical understanding of loading conditions during many strength exercises remains incomplete, most guidelines and recommendations on different strength exercises have to resort to subjective experience of individual experts, coaches, and athletes.

1.1.1 Sports performance

Sport is a truly an omnipresent part of our every-day life, influencing nearly every aspect of our society, from culture, over health, to economy. In the United States, 44 million children and adolescents (63% boys and 37% girls) participated in various organized sport activities in 2008 [18]. In Switzerland, with an increasing interest on sports activity, only one quarter of all surveyed persons (aged between 15 and 74 years) stated not participating in sports at all. In the same study, “health” and “fun” were given as main reasons to perform sport on a general basis, and “performance enhancement” and “sociality” predominately chosen by younger athletes [1].

Due to the fact that nearly every sport requires a certain amount of muscle resistance and power, strength training has become a key aspect in any athletes’ training program to enhance muscle strength, increase muscle cross-sectional area, and improve muscle coordination (inter- and intramuscular coordination) [19]. This especially includes any sport in which sprint, jumping, or muscle power ability play a considerable role, as these abilities have been shown to relate to particular muscular strengths and thus can be improved by specific strength training [20, 21].

It is recommended to increase force-generation capacities of sports related important muscles to enhance sport performance [19]. Because different sports require different muscles and muscle activation patterns, successful strength training must identify the specific muscles as well as offer an efficient way to train these muscles in the appropriate way. Furthermore, the athlete’s physical condition (age, experience, fitness level, strength, medical conditions, etc.) need to be
taken into account when developing athlete specific training designs [19–21].

While past research has investigated the biological mechanisms underlying the muscle adaption and the general health benefits related to strength training, open questions remain on how to identify the muscular demands for a certain sport activity and how to train these structures in an efficient, safe, and sustainable manner.

1.1.2 Prevention and Rehabilitation

Strength training has the benefit of strength and power enhancement and is common used in prevention programs with the aim to reduce the risk of injury in other sport activities [22–24]. For example, the risk for ACL tears can reduced by 88% [24] by performing an sports-specific training intervention including strength training.

Apart from the suffering on an individual level, injuries and the resulting healing period also cause a long recovery phase and decrease in sport performance as well as high medical costs. In Switzerland, approximately 8% of the population sustain a sport injury each year, especially during soccer and skiing [1]. In both soccer and skiing, the knee joint is the most likely joint to be injured, constituting 27% [25] and 60% [22] of all sustained injuries, respectively. In such knee injuries, the anterior cruciate ligament (ACL) is one of the most commonly injured ligaments with an estimated 400’000 ACL tears per year in the USA [26] and 10’000–12’000 tears per year in Switzerland, which correspond to 0.12% and 0.14% of the population. Highest numbers of ACL injuries occur through rapid but awkward stops, lack of anticipation of lateral movements, and landing with a femoral adduction and internal rotation, creating a valgus knee [27]. Interestingly, it seems that women are more likely to suffer from an ACL rupture [28]. ACL tears can be treated using either both an ACL surgical reconstruction combined with a structured rehabilitation or using a structured rehabilitation only. Mean lifetime costs to society are estimated as $38,000 for patient undergoing ACL reconstruction compared to $88,538 per patient undergoing structured rehabilitation only [29]. This cost indicates the necessity to reduce the risk of injury of ACL tears, but also that structured rehabilitation can still be improved.

Successful recovery following a procedure or injury depends critically on appropriate loading but the treated tissue must not be overloaded [4–7]. Even though there are high chances of a patient recovering from a sport injury, the necessary treatments are often described insufficiently [30–33], so recovery success depends mainly on the experience of the attending physician or the
physiotherapist [4, 7, 8, 34, 35]. This lack of detailed description, for example on the necessary step length while performing split squats to recover from patellofemoral pain syndrome [36] leads to unnecessary long rehabilitation or even injury recurrence during the rehabilitation process [4]. A currently accepted way to reduce the risk of ACL rupture is to increase the force of the hamstrings compared to the quadriceps force (H:Q-ratio), since balanced co-contraction enhances the stabilisation of the knee and therefore prevents the anterior translation of the tibia with respect to the femur [23, 37–39]. Beside the H:Q-ratio, knee stabilisation seems to play another key factor to reduce the risk of ACL ruptures. Since ALC tears mainly occur during the early phase of an impact movement such as during landing, stabilisation of the knee in this particular phase needs to be guaranteed. Activation of the hamstrings in this time period leads to an enhanced stabilisation and is an important factor to reduce the risk of ACL ruptures. The recommended way to enhance this activation level is to strengthen hamstring musculature using a combination of high intensity strength training [40], balance training [41], and proprioceptive training [42], all in a sport-specific manner. To train in such a specific manner, first the sport’s characteristic loading and movement patterns have to be established, followed by an appropriate choice of strength exercise to address these requirements [43].

Thus, performing strength training is state of the art to enhance sports performance, to reduce risk of injury, to recover from an injury or surgery, as well as to decrease medical costs.

1.2 Internal loading conditions

Knowledge about the internal loading conditions arising during strength exercises is necessary to develop both general recommendations and individualized training and rehabilitation programs. Unfortunately, these internal loading conditions remain unknown for most strength exercises, due to the fact that current techniques to measure the internal loading conditions in vivo are highly limited. While no measurement technique allows in vivo muscle force measurements, direct measurement of joint contact forces in vivo is possible but restricted to elderly subjects possessing an instrumented arthroplasty. Knowledge on muscle forces is of high interest, since the aim of performing strength training is to strengthen the muscles, but also due to the fact that loading conditions are predominantly affected by these forces.

As an alternative to in vivo measurements, internal loading conditions can be calculated indirectly from surface measurements, such as EMG, dynamometer, or inverse dynamics. These
calculations are complicated by the over-determination of the system (e.g., multiple muscles crossing one joint; Figure 1.1) and can thus only yield approximate results. Using data recorded from dynamometers or using an inverse dynamic approach, calculations of external joint moments are possible only. External moments neglect the acting internal forces, intra- and intermuscular coordination, and co-contraction [44]. Internal forces are known to change loading conditions in a complex manner and should therefore not be neglected. On the other hand, surface EMG provides a better understanding on the muscle activation levels, but EMG-force relationship is known to be sensitive to dynamic movements. Thus, estimations of muscle forces using surface EMG is possible in static positions only [45, 46]. None of these above mentioned methods provide the possibility to estimate internal loading conditions in a non-invasive, accurate way.

Figure 1.1: Simple model to show mechanically over-determined system of muscles. Since the number of muscles crossing a joint is greater than the number of degrees of freedom of the joint, optimization criteria are required (Figure provided by J. Denoth).
1.3 Musculoskeletal Simulation

Knowledge on internal loading conditions is essential to provide an efficient and safe training to athletes or patients. None of the above mentioned methods (Chapter 1.2) provide the possibility to estimate internal loading conditions in a non-invasive, accurate way. This problem can be addressed through the use of optimization techniques. Since the number of muscles crossing a joint is greater than the number of degrees of freedom at the joint, optimization techniques and the choice of appropriate optimization criteria are required to determine internal loading conditions. Computational musculoskeletal (MS) numerical simulations or other modelling techniques therefore provide a link between measured data and internal loading conditions during exercises (Figure 1.2).

Figure 1.2: Simulation software Biomechanics of Bodies [47] used to model joint contact forces, joint moments and muscle forces during strength exercises split squats.

MS simulation software has been developed to estimate the internal loading conditions from externally measured data. In order to correctly predict muscle and joint reaction forces, a MS model has to consider the subject’s anatomy, the force generating tissues as well as the performed movement. By imitating kinematic and kinetic data of the movement, a MS model relates the movement to the internal loading conditions through Newton’s laws of motion [48]. Since the system is over-determined (Figure 1.1), the usage of a minimization criterion is required
to find the unique solution for the muscle, joint reaction, and passive forces occurring during
the movement. Various commercial software packages are available to calculate internal loading
conditions from externally measured data [47, 49].

Computational MS models have been widely used in clinics, for example to plan joint replacement
[50, 51] and its outcomes [52], or to estimate loading conditions in sports and daily activities
[53, 54]. MS simulation results are highly dependent on implementation details, amongst others,
the inclusion of:

- Subject-specific scaling
- Muscle activation criteria such as force-length-velocity dependence
- Co-contraction
- Correction of soft tissue artefacts
- Passive forces applied by ligaments, as well as forces based on soft tissue compression
- optimization criteria (also called cost functions)

Although recent developments in the underlying MS simulation techniques have been made,
confirmation of the efficacy of these changes remains challenging, especially during strength
training. Therefore, further investigation is needed to confirm the applicability and usefulness
of MS simulations for real life applications.
1.4 Conclusion

Strength training has the ability to enhance sport performance but also to prevent injuries as well as support rehabilitation from already sustained injuries such as ACL rupture and thus improving people’s quality of life. Loading conditions but also joint angles seem to be key factors to ensure a safe and efficient adaptation of a tissue. Here, a certain amount of loading is required to induce a positive adaptation of muscle structures - however, care has to be taken because excessive loading could lead to injury or inflammation. External kinematics and kinetics during strength exercises are known to influence the internal forces and the adaptation of biological structures in a complex manner. Knowledge about these internal loading conditions could allow efficient and evidence-based development of training strategies, rehabilitation programs, as well as general health recommendations and thus reduce the risk of injury not only in strength exercising itself but also in other sports. The aims in an ACL rupture rehabilitation process are to enhance the stability as well as to strengthen the muscles surrounding the knee. To ensure an adaptation and avoid overloading, specific loading conditions are crucial and therefore, appropriate training designs are required.

Due to the fact that direct measurements of these loading conditions are not possible with current techniques, modelling approaches such as quasi-static inverse dynamic approaches or musculoskeletal simulations allow these external and internal loading conditions to be accessed [55, 56]. The outcomes of these modelling approaches are highly dependent upon input factors. Therefore, robust and accurate simulation techniques as well as real guidelines have to be developed and identified to provide estimated loading conditions including muscle forces during strength exercises. Both loading conditions during strength exercises as well as specific training designs for ACL rehabilitations remain elusive in literature. It is therefore critical to better understand the complex relationship between the external movement and loading patterns, including exercise type, on the internal loading conditions within the soft tissue structures surrounding the knee.

1.5 Specific Aims

Arising internal loading conditions, including specific muscle or passive forces, during strength exercises are essential to achieve a fast, safe, and focussed adaption. These loading determine whether a fast, safe, and efficient tissue adaption can take place, or whether no effect or even damage occurs in the tissue. Therefore, with the aim to improve prevention guidelines, reha-
bilitation programs, as well as training strategies to improve sports performance, this doctoral thesis assesses:

- The different possibilities to assess loading conditions during strength training.
- The accurate representation of measured motion data in a musculoskeletal model using functionally defined centres of rotation and axes of rotation.
- The validity of the calculated loading conditions through comparison with the measured joint contact forces during a strength exercise.
- The external moments and corresponding joint angles during different strength exercises.
- The estimated muscle forces during different strength exercises.
- Based on estimated loading conditions, the recommendations for training, prevention, and rehabilitation programs to reduce risk of ACL tears.

1.6 Outline of the thesis

The thesis is divided into 8 chapters. Each of the six Chapters 2, 3, 4, 5, 6 & 7 are presented based on peer-reviewed scientific article, already published or submitted to an international scientific journal.

Chapter 1 provides a short introduction on the topic, motivation, and background of the project and presents the specific aims as well as the outline of this thesis.

Chapter 2 provides different approaches to estimate loading conditions, with a focus on muscle forces. A literature review on the modelling techniques for in vivo muscle force estimation in the lower extremities during strength training are presented.

Through an identified approach in chapter 2 (quasi-static inverse dynamic), chapter 3 and chapter 4 investigate in loading conditions and movement patterns arising during deadlift, goodmornings, split squats and hyper-extension exercises. Specific training and rehabilitation recommendations, with a special focus on ACL ruptures, are developed from the results.

Results gained from chapters 3 & 4 provide important and helpful recommendations, however, internal loading conditions seem to be more meaningful and have the ability to provide more precise training designs and rehabilitation recommendations. Since the outcome of musculoskeletal
simulations is highly dependent upon simulation techniques and input factors, chapter 5 investigates on the robustness of different simulation weighting concepts and on the reduction of the error between reference motion data and the musculoskeletal simulation. The results allow real guidelines for registering a generic musculoskeletal model (OpenSim) to reference kinematic and kinetic data for the individualisation of simulations to be developed.

Using the developed simulation technique in chapter 5 to represent motion data in a robust and accurate way, chapter 6 evaluates the error of calculated tibio-femoral joint contact forces compared to in vivo measured joint contact forces using instrumented total knee arthroplasties of 6 subjects during squat strength exercises. The validation of the standard model allows accurate interpretation of the estimated muscle forces.

The validated musculoskeletal simulation technique in chapter 6 provides the ability to estimate muscle forces during strength exercises. Chapter 7 therefore investigates the calculated muscle forces acting in the lower limb muscles during the strength exercises deadlifts, goodmornings, and split squats using individualised models and developed simulation technique. Training, prevention, and rehabilitation recommendations, especially for ACL ruptures, are designed and presented.

Chapter 8 summarises the key findings and outcomes of all the previous chapters and provides an in-depth discussion of their limitations, outcomes, and implications. Furthermore the collective results are discussed with regards to literature and a brief outline of further developments and possible follow-up studies is presented.
Chapter 2

Review of Modelling Techniques for *In Vivo* Muscle Force Estimation in the Lower Extremities during Strength Training

adapted from:

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open access
2.1 Abstract

2.1.1 Background

Knowledge of the musculoskeletal loading conditions during strength training is essential for performance monitoring, injury prevention, rehabilitation and training design. However, measuring muscle forces during exercise performance as a primary determinant of training efficacy and safety has remained challenging.

2.1.2 Methods

In this manuscript we review existing computational techniques to determine muscle forces in the lower limbs during strength exercises in vivo and discuss their potential for uptake into sports training and rehabilitation.

2.1.3 Results

Muscle forces during exercise performance have almost exclusively been analysed using so-called forward dynamics simulations, inverse dynamics techniques or alternative methods. Musculoskeletal models based on forward dynamics analyses have led to considerable new insights into muscular coordination, strength and power during dynamic ballistic movement activities, resulting in, for example, improved techniques for optimal performance of the squat jump; while quasi-static inverse dynamics optimisation and EMG-driven modelling have helped to provide an understanding of low-speed exercises.

2.1.4 Conclusion

The present review introduces the different computational techniques and outlines their advantages and disadvantages for the informed usage by non-experts. With sufficient validation and widespread application, muscle force calculations during strength exercises in vivo are expected to provide biomechanically-based evidence for clinicians and therapists to evaluate and improve training guidelines.
Keywords

Resistance training; musculoskeletal modelling; inverse dynamics; forward dynamics; computational biomechanics; sports
2.2 Introduction

The quantification of muscle forces during muscle strengthening exercises \textit{in vivo} has tremendous potential for assisting with training design, performance monitoring and injury prevention [57]. Due to the fact that 45.6\% of all injuries during strength training in Switzerland occur due to overloading [14], and current exercise guidelines as well as training recommendations are based on the subjective experience of individual experts or coaches, knowledge about the effects of external loading on internal muscle forces during training or rehabilitation could help to improve exercise safety. In addition, the analysis of the internal loading conditions provides an evidence-based approach for defining specific targets and loading goals for effective training outcome while also reducing injury risk, since fewer loads can be used to achieve the same training effect. Furthermore, muscles are able to specifically adapt to their loading output to the surrounding functional requirements and must therefore respond appropriately to allow the rehabilitation of unbalanced musculature in an effective and safe way. However, measuring muscle forces \textit{in vivo} remains challenging due to the complexity of movement control, the non-linear material properties of muscle tissue, the redundant number of muscle actuators and the invasive nature of direct measurement techniques [58]. Existing guidelines on strength training (type of exercise, repetitions, number of sets etc.) are often based on experience or simple measurements from dynamometry or surface electromyography (EMG) [44, 46, 59, 60]; but the actual stress levels in the muscle based on muscle force and cross-sectional area measures, which provide direct evidence for the efficacy and safety of specific muscle-strengthening exercises, have been difficult to obtain [16, 17].

It is not currently possible to measure muscle forces experimentally during exercise performance \textit{in vivo}, and data from alternative measurement techniques have not been sufficient for deducing internal forces and moments for complex dynamic systems, such as the lower limbs, in a straightforward manner [49]. Common measurement techniques in human motion analysis and sports science include surface EMG, optical motion capture and force measures from e.g. dynamometer or force platforms [16]. Dynamometry has frequently been used to determine strength and power during open chain leg extension or flexion exercises [44]. However, strength and power are scalar variables that provide only limited insights into actual muscle forces that occur internally, especially considering the complexity of inter- and intra-muscular activity and coordination associated with free weights or multi-joint dynamic training [44]. On the other hand, surface EMG provides a good insight into muscle activation levels during functional exercise performance compared
to strength and power measures gained from dynamometry. However, while EMG offers more specific information on muscle function compared to dynamometry, it still provides insufficient data to deduce muscle force magnitudes [46], especially when analysing dynamic movements [45].

Computational models of the musculoskeletal system are therefore needed to provide a link between externally measured data and internal forces and moments (Figure 2.1). Musculoskeletal modelling techniques have been developed and extensively used in clinical and biomechanical gait analysis, in particular for studying lower limb dynamics. In order to predict muscle forces during movement, a computational model has to capture the anatomy of the musculoskeletal system, as well as the physiological, force generating properties of muscle tissue, and then relate the target movement to the internal muscle forces through Newton’s laws of motion [48]. Additional parameters, such as individual ratios of fast-twitch versus slow-twitch fibres within each muscle or muscle versus fat volumes within the segments, could be taken into account in an optimization process and enhance the accuracy of a model but also its complexity (Figure 2.1).

Figure 2.1: Muscle and joint forces are quantified in vivo by combining experimental measurements (yellow) with computational biomechanics (orange). Different measurement parameters (black arrows) or computational optimizations (black arrows) are required to achieve different output parameters (green) in inverse dynamics or forward dynamics processes. For forward dynamics simulations (red arrows), usually applied to dynamic ballistic movement exercises such as the squat jump, joint dynamics such as joint angles, joint net moment, or muscle kinematics are derived by finding an optimal set of muscle kinetics using computational modelling. For inverse dynamics analysis (blue arrows), usually applied to low-speed exercises such as the squat, joint moments, muscle forces, and finally joint contact forces are derived from joint angles and net joint moments.

Depending on model complexity, available experimental data and study goal, the dynamic sys-
tem of equations (originating from Newton’s second law of motion) can be solved in different ways, including forward dynamics \([61–68]\), inverse dynamics \([15, 46, 48, 69–71]\) and EMG-driven analyses \([59, 60, 72, 73]\). Using forward dynamics, a set of muscle activation patterns are usually chosen as input into a physiological muscle model to derive muscle forces. Muscle forces are then applied to a rigid body skeletal model to estimate joint moments or joint angles (Figure 2.1). In contrast, inverse dynamics uses data from experimental measurements including skin marker positions and ground reaction forces as input into a rigid body skeletal model to calculate joint net moments. In addition, using optimization processes and musculoskeletal modelling, joint forces and moments can be computed from the results of the inverse dynamics analysis. An EMG-driven analysis uses normed muscle activation levels from EMG measurements in addition to skin marker positions and ground reaction forces to improve the estimation of muscle force magnitudes by means of musculoskeletal modelling. Unfortunately, the unknown muscle forces that cause a particular movement generally exceed the known parameters from experimental measurements, resulting in redundant systems of equations that require the use of various optimization techniques \([59, 61, 68, 71, 73]\) (Figure 2.1).

Improved knowledge of the specific muscle forces that act during strength training could help coaches and athletes improve training protocols, as well as physiotherapists and patients to undertake rehabilitation exercises in an efficient and safe manner. Furthermore knowledge of the muscle forces can be used as boundary conditions within continuum organ models in order to estimate the biological change of the tissues such as muscle, bone, tendon due to the mechanical stimuli of strength training. However, current approaches for deriving muscle forces are generally complex and require substantial expertise in computational modelling. In this review of the literature, we introduce non-experts to current musculoskeletal modelling techniques for determining muscle forces (generally of the lower limbs) during strength training \(in vivo\), and discuss their potential as well as limitations for application to sports practice in order to assist with training recommendations and guidelines. In this manner, this review aims to result in an improved understanding of the existing computational techniques, and thus provide a basis for future developments and more widespread and informed application of available biomechanical tools. Eventually, the results of in-depth biomechanical analyses are expected to help define objective, evidence-based guidelines for coaches and therapists to execute strength training exercises in an effective and safe manner.
Chapter 2: Review of Modelling Techniques for In Vivo Muscle Force Estimation

2.3 Methods

A systematic electronic literature search of the US National Library of Medicine was conducted in August 2013. Four combined concepts including different search parameters were used to systematically search the literature (Table 2.1). A total of 77 papers were found complying with all four concepts and were further assessed for eligibility to be included in the review based on the following exclusion criteria: 1) no measurements were taken during a functional strength exercise of the lower extremities, or 2) results were limited to EMG data, maximum voluntary isometric contraction or net joint moments without adopting a computational model to determine muscle forces, or 3) ex vivo study. Out of these 77 papers, a total of 12 articles were eligible for the full review. Articles were mainly excluded because results were limited to measured data from EMG or dynamometry, without adopting musculoskeletal modelling techniques to determine muscle forces or other internal forces. Additional studies were excluded because muscle forces were analysed during activities other than strength training or were ex vivo, performed on cadaveric specimens. Strength training was defined as a physical exercise that induces muscular contraction to enhance strength, anaerobic endurance or size of a skeletal muscle. References from the included papers were further searched for relevant work including the same criteria. An additional 9 papers were found within the references and included in the review process. Ultimately, the full text articles of 21 studies were included in the review.

Table 2.1: Four concepts of search parameters were used to systematically search the literature and were combined using an "and" condition (horizontal). Each concept was created using "or" conditions (vertical) in order to ensure the inclusion of all papers using similar definition for the same case.

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*They were used as wildcards to replace part of a string.
Chapter 2: Review of Modelling Techniques for *In Vivo* Muscle Force Estimation

2.4 Results

Musculoskeletal models with different levels of anatomical detail and computational complexity have been developed to determine muscle forces during strength exercises of the lower extremities *in vivo* (Table 2.2). The 21 included studies were divided into three categories, with 9 studies performing forward dynamics simulation, 2 studies adopting quasi-static-inverse dynamics optimisation, and 10 studies outlining mixed inverse/forward dynamics (1 study) or mixed inverse dynamics/alternative methods including EMG-driven modelling (9 studies) (Table 2.2). Forward dynamics simulations have predominantly been used to study dynamic ballistic movement exercises such as the squat jump; while quasi-static inverse dynamics optimisation techniques or alternative methods have been adopted to analyse low-speed exercises such as the lunge or leg press under the assumption of "negligible acceleration in each segment". Although Pearsall and Costigan [77] showed that accurate net moments may be achieved using quasi-static approaches in low-speed exercises, some studies have used a true inverse dynamics approach with inclusion of the accelerations of the segments and their moments of inertia [78–81]. All methodologies have in common that they draw on the physics principles of multi-body dynamics (Newton’s laws of motion) and rely on accurate representations of musculoskeletal anatomy and physiology to accurately predict the muscle forces that cause the target motion (Figure 2.1).

2.4.1 Anatomical and physiological model parameters

Musculoskeletal models with different numbers of muscles and joints, and different material property characteristics for active muscle and passive soft tissues, have been introduced depending on the research goal, available data and expertise in computational modelling. Musculoskeletal models with a reduced number of muscles and/or a lower number of degrees of freedom at the joints have been adopted to simplify analyses. In particular, the knee joint has generally been represented as a planar hinge joint, neglecting translational and rotational degrees of freedom other than flexion-extension [46, 48, 61, 65, 68, 69, 71, 73, 75, 76]. In these models, individual muscles have often been grouped to reduce the unknown degrees of freedom of a musculoskeletal model, such as the hamstrings or quadriceps [65, 68, 69, 71, 72], or only key muscle players have been considered [46, 74]. Exceptions are the anatomical models adopted by different authors [48, 75, 76], which considered up to 43 muscle-tendon units per leg. These authors additionally subdivided large or complex muscles such as the gluteal muscles into multiple muscle units to more accurately represent their muscle paths and functions than would single muscle units.
### Table 2.2: Summary of studies reporting on computational techniques to determine muscle forces during strength training of the lower extremities in vivo.

Dynamic squat jumps were mainly analysed using forward dynamic (FD) simulation, while low-speed ankle, hip, and knee exercises were analysed using quasi-static inverse dynamics (ID) optimisation, electromyography-driven (EMG) modelling, or mixed inverse dynamics/forward dynamics analysis. Different approaches were adopted to distribute the net joint moments from ID across muscles, ranging from simple 1-muscle models to advanced optimization schemes taking muscle force-length-velocity (F-L-V) into account. Data from EMG, optical motion capture (OMC), and ground reaction forces (GRF) were used as input or reference to assess the accuracy of modelling results.

<table>
<thead>
<tr>
<th>Exercise</th>
<th>Modelling approach</th>
<th>Subjects</th>
<th>Exp. measure</th>
<th>Reported results</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low-speed foot plantar flexion</td>
<td>ID (1-muscle model)</td>
<td>8M, 8F (22y)</td>
<td>EMG, OMC, GRF</td>
<td>Muscle force</td>
<td>[46]</td>
</tr>
<tr>
<td>Low-speed deep knee bends</td>
<td>ID (1-muscle model)</td>
<td>3M (26y)</td>
<td>OMC, GRF</td>
<td>Muscle and joint forces</td>
<td>[74]</td>
</tr>
<tr>
<td>Low-speed squat, leg press</td>
<td>ID (F-EMG)</td>
<td>10M (30y)</td>
<td>EMG, OMC, GRF</td>
<td>Tibiofem joint kinetics, cruciate ligament force</td>
<td>[15, 69, 70]</td>
</tr>
<tr>
<td>Low-speed squat, leg press</td>
<td>ID (optimised F-L-EMG)</td>
<td>9M (29y), 9F (25y)</td>
<td>EMG, OMC, GRF</td>
<td>Patellofemoral force and stress</td>
<td>[59, 60]</td>
</tr>
<tr>
<td>Low-speed squat, leg press</td>
<td>ID (optimised F-L-V-EMG)</td>
<td>9M (29y), 9F (25y)</td>
<td>EMG, OMC, GRF</td>
<td>Patellofemoral force and stress</td>
<td>[59, 60]</td>
</tr>
<tr>
<td>Low-speed hip extension/flexion</td>
<td>ID (min stress)</td>
<td>Generic simulation - Hip joint forces</td>
<td>[48]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Low-speed abdominal crunch</td>
<td>Mixed ID/FD equipment</td>
<td>Generic simulation - Intervertebral joint loading</td>
<td>[75]</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>6M (25y), well-trained volleyball players</td>
<td>EMG, OMC, GRF</td>
<td>Gastro biarticularity</td>
<td>[62]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>6M (25y), well-trained gymnastics</td>
<td>EMG, OMC, GRF</td>
<td>Muscle strengthening</td>
<td>[61]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>6M (25y), well-trained volleyball players</td>
<td>EMG, OMC, GRF</td>
<td>Tripeps strain series elastic compliance</td>
<td>[62]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>6M (25y), well-trained volleyball players</td>
<td>EMG, OMC, GRF</td>
<td>Fatigue of muscles</td>
<td>[63]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>5M (22y)</td>
<td>EMG, OMC, GRF</td>
<td>Bilateral deficit</td>
<td>[76]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>5M (22y)</td>
<td>EMG, OMC, GRF</td>
<td>Optimal control</td>
<td>[77]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>5M (22y)</td>
<td>EMG, OMC, GRF</td>
<td>Contributions of muscles to accelerate trunk</td>
<td>[78]</td>
</tr>
<tr>
<td>Dynamic ballistic squat jump</td>
<td>FD (Activation <em>A</em> = 0.1)</td>
<td>5M (22y)</td>
<td>EMG, OMC, GRF</td>
<td>Bilateral asymmetry</td>
<td>[79]</td>
</tr>
</tbody>
</table>
The anatomical and physiological properties of these three models were based on experimental measurements of cadaveric specimens from the literature in an attempt to represent the modelled subject-specific properties [76]. Two of the models were generically implemented into commercially available or open-source software packages (LifeModeler™, OpenSim) [48, 75].

The active and passive material properties of muscle-tendon structures have commonly been described by a Hill-type [82] active element, which comprises a so-called contractile element (CE) (force generation by actin and myosin cross-bridges) to capture the force-length-velocity dependency of muscle tissue. This active element is additionally coupled with two passive elements (represented by non-linear spring elements); a series-elastic (SE) element to account for the tendon elasticity and a parallel-elastic element (PE) to account for the passive stiffness of the muscle connective tissue (Figure 2.2) [61–68, 76]. The force-velocity dependency of a particular muscle can thus be derived directly from Hill’s equation $(v+b)(F+a) = b(F_0+a)$ [82], where larger forces can be produced by the CE during slow velocity contractions and vice versa. During eccentric movement, muscles are able to produce even higher forces, since passive structures additionally support the CE for force production. The force that can be produced by a muscle is further dependent on the actual length of a muscle, since the force generated actin myosin cross-bridges at the sarcomere level are depend on their overlapping status. Furthermore passive structures (PE and SE) produce force when the muscle is stretched, even if the CE is not activated. The Hill-type muscle model has widely been accepted by the biomechanics community and has been implemented into musculoskeletal modelling software packages such as OpenSim [49, 83].

![Diagram of the Hill-type muscle-tendon model](image)

**Figure 2.2**: The Hill-type muscle-tendon model, showing muscle and tendon forces ($F^M$, $F^T$), as well as the series-elastic (SE), parallel elastic (PE), and contractile (CE) elements of the muscle length ($l$) and stiffness ($k$) of the whole muscle-tendon actuator ($M$, $T$). $a(t)$ represents the activation of the CE (adapted from Pandy and co-workers [61]).

Material parameters (pre-determined, constant values) for the Hill-type muscle model are gener-
ally derived from experimental measurements on cadaveric specimens reported in the literature, including maximum isometric force, muscle fibre pennation angle, tendon slack length, tendon and muscle passive stiffness and physiological cross sectional area (PCSA). Simplifications to the Hill-type model have been made by neglecting the force-velocity and/or force-length relationship as well as the passive material properties, especially to analyse quasi-static exercises using inverse dynamics techniques [15, 48, 69–71, 75]. Adjustment of material parameters to individual subjects is generally achieved through simple scaling based on segmental lengths, calculated joint centers, EMG signals (normalized to the maximal voluntary isometric contraction (MVIC)), or subjects body weight [15, 59–61, 66–71]. Alternatively, static or functional optimisation approaches can be used [73]. Other techniques to determine individual muscle parameters include ultrasound measurements, which allow the evaluation of muscle volume and thus PCSA [58], but no studies have been found that combine ultrasound measurements with musculoskeletal modelling to analyse muscle forces during strength training.

2.4.2 Forward dynamics simulation

In general, the problem to simulate a musculoskeletal model using a forward dynamics approach is to find a physiologically feasible set of controls regarding the muscle activity, for example, by means of the minimized integral cost function. This usually includes a large set of boundary conditions and constraints to be defined. A set is related to different application area such as movement, loading conditions, physiology and the time dependency of an optimization algorithm. Furthermore, forward dynamics simulations are usually associated to a control problem. Here, open loop solutions are often highly unstable and difficult to integrate while close loop solutions of the highly nonlinear musculoskeletal system remain still an unsolved problem. Significant attempts have been made to simulate the musculoskeletal system in motion based on forward dynamics, using muscle activation levels as input and the time-history of segmental positions and orientations as output, in particular to study dynamic ballistic movement exercises such as the squat jump. Here, different studies have introduced forward dynamics models of the musculoskeletal system to better understand, for example, how inter-muscular control [61, 62, 66], bilateral-asymmetry [63, 76] or muscular fatigue [64] affect the maximum jump height. The dynamic equations of motion of the skeletal system are thereafter described by a set of differential equations, driven by muscle-tendon actuators which are controlled by neural signalling. Muscle-tendon actuators are commonly represented using the Hill-type muscle-tendon model, connected to a numerical model to capture the time lapse between the incoming neural signal and the
onset of muscle activation. Forward dynamics simulations depend on optimisation algorithms to find feasible sets of muscle activation patterns leading to the desired movement dynamics, with maximum jump height as a common performance criterion. For other movements such as squats or lunges, new criteria would need to be defined. Solutions to the optimisation problem are more likely found for unambiguous movement patterns with simplified musculoskeletal models, i.e. with a limited number of muscle actuators and constrained conditions such as reduced degrees of freedom of the joints [61].

One of the first forward dynamics models for the planar squat jump was introduced by Pandy and co-workers [61], comprising all lower limb bones and eight major muscles (Figure2.3). The constraints that defined the optimal control problem were the dynamic equations of motion, the terminal calculation point at take-off and the muscle activation levels being set to 0 or 1, with maximum jump height as the performance criterion. A more restricted form of dynamic optimisation was introduced by van Soest and co-workers [68], whereby muscles were only allowed to switch from the initial activation values once, and then had to maintain maximum activation until take-off. The problem was thus reduced to finding an optimal combination of the muscular switching times to result in maximum jump height. The resulting muscle activation patterns corresponded well with experimentally measured data from EMG, and the formulation has often been applied to biomechanical analyses of the squat jump [62–66].

In order to gain confidence in forward dynamics simulations, modelling results have often been compared with experimental data from optical motion capture, force platforms and EMG [63–65, 67, 68], confirming the ability of forward dynamics models to accurately reproduce the major features of maximum-height squat jumps. Based on the present literature search, and in agreement with previous reviews on muscle force calculations in orthopaedics and clinical gait analysis [83], forward dynamic methods have not yet been applied to muscle strengthening exercises other than the squat jump. While the performance criterion for the squat jump is generally maximal jump height, the selection of performance criteria for other activities is considered more challenging [83]. Furthermore, forward dynamic models require multiple integration steps to reach optimal joint kinematics, resulting in a computational complexity that limits their implementation in user-friendly software packages, and thus, widespread use by non-experts. However, using forward dynamics offers the possibility for coaches or therapists to simulate an optimal training or rehabilitation program for a specific athlete or patient without the need for elaborate experimental measurements such as EMG, optical motion capture, or ground reaction forces.
Figure 2.3: (a) Schematic representation of the musculoskeletal model for the vertical jump and (b) the four-segment multibody model with lumped masses and mass moments of inertia for the foot, shank, thigh, and head/arms/trunk (Pandy and co-workers [61]).
Quasi-static inverse dynamics optimisation

In contrast to forward dynamics simulation, the inverse dynamics formulation is comparably quick and computationally inexpensive. Inverse dynamics analysis refers to the calculation of segmental forces and moments based on data from optical motion capture and force sensors such as force platforms, and has become a routine tool in clinical gait analysis [83] and strength training exercises [84–87]. It is important to note that joint contact forces and muscle forces cannot be calculated directly from inverse dynamics. Instead, the derivation of muscle forces necessitates distributing the net inter-segmental forces from inverse dynamics across synergistic and antagonistic muscles, which leads to a problem of indeterminate nature that needs to be solved using numerical optimisation techniques. Joint contact forces can additionally be calculated as the sum of the net inter-segmental forces and the synergistic and antagonistic muscle forces that cross the joint (Figure 2.1).

Quasi-static inverse dynamics optimisation techniques have generally been applied to low speed movements, such as the leg press or lunge [15, 46, 48, 69, 71, 74]. For low-speed exercises, the quasi-static equilibrium condition holds true under the assumption that the angular acceleration of each segment is negligible. In an early study, Reilly and Martens [74] introduced a single-muscle model to quantify quadriceps muscle forces during deep knee bends based on inverse dynamics. The model worked under the assumption that only the quadriceps as a single muscle group is active during the exercise. Thus, the net knee joint moments from inverse dynamics were equal to the resulting muscle moments, and muscle forces could be determined geometrically by deriving the moment arms with respect to the knee joint centre. A similar modelling approach was adopted by Henriksen and co-workers [46], analysing the eccentric and concentric forces in the Achilles tendon during ankle plantar- and dorsi-flexion. Here, the advantage is that single-muscle models do not depend on computationally expensive optimisation techniques; however, the potential contribution of synergistic and antagonistic muscles to joint stability and movement control is neglected and the physiological differences in force generating capabilities between muscles cannot be accounted for.

More complex musculoskeletal models based on quasi-static inverse dynamics have been developed, accounting for the contribution of synergistic and antagonistic muscle groups to analyse open and closed chain knee extension [15, 69, 70] and hip flexion-extension [48]. Here, optimisation algorithms based on the least-squares method have generally been adopted to find weighting factors for each muscle force contribution to minimize the differences between the inter-segmental
torques from inverse dynamics and the resultant muscle torques from the biomechanical model. In early attempts, muscle forces were assumed to be proportional to physiological cross-sectional area (PCSA), maximum voluntary contraction force and measured EMG activation levels, without taking the force-length [71], muscle fibre recruitment [46] and force-velocity [69, 70] relationships into account. To improve results, Zheng and co-workers [71] extended previous models by examining the role of muscle force-length properties, and demonstrated that force-length dependent optimisation during squat and leg press exercises had a significant effect on muscle force magnitudes and proved an important factor in determining tension in the cruciate ligaments.

A slightly different approach based on quasi-static inverse dynamics optimisation was adopted by Lewis and co-workers [48] to analyse the effect of position and alteration in synergist muscle forces on hip forces during hip strengthening exercises. Muscle-tendon paths and maximum isometric forces of 43 muscle units were adopted from a generic musculoskeletal model in the commercially available software SIMM (MusculoGraphics, Inc, Santa Rosa, CA, USA). Muscle material properties other than muscle-tendon paths and maximum isometric force were neglected, including force-length relationships and the passive response to stress. An optimisation algorithm was adopted that aimed to minimise muscle stress with the goal of maximizing muscle endurance. Such an approach has been widely accepted for biomechanical analyses of the lower limbs during gait [83]. However, quasi-static optimisation techniques based on minimizing muscle stresses have been shown to underestimate antagonistic muscle activity as well as muscle force contributions of low magnitudes [88]. Furthermore, subjects who are fatigued or in pain are unlikely to activate muscles according to a minimal effort principle but rather with the aim of avoiding mechanical stress on fatigued or painful tissue.

2.4.4 Alternative Methods

A group of alternative methods have been introduced to calculate muscle forces based on a mixed inverse-forward dynamics approach [75] or by using EMG data to drive a musculoskeletal model towards given joint kinematics (EMG-driven modelling) [59, 72, 73]. In particular, Nolte and co-workers [75] outlined a combined inverse-forward dynamics simulation to quantify intervertebral loading during the abdominal crunch exercise based on a full-body musculoskeletal model using the LifeModeler™ software. The modelling output depended upon the initial estimation of muscle-tendon lengths from inverse dynamics to provide reference values for the derivation of muscle activation levels, and thus muscle forces. However, the model did not account for
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Physiologically realistic material properties of muscle tissue. Instead, muscle forces were derived using a closed loop algorithm containing proportional-integral-differential controllers to reach the target length-time curve. Despite questions arising in terms of model validity, the study remains unique in that it included a Computer Aided Design (CAD) model of the training machine. By using musculoskeletal models of different sizes together with the CAD model, the authors were able to analyse the effectiveness and safety of the exercise machine to accommodate very small or large individuals based on the predicted muscle forces and intervertebral joint loading.

Other alternative methods to determine muscle forces during strength training include so-called EMG-driven musculoskeletal models, introduced by Lloyd and Besier [73] and Escamilla and co-workers [59, 60]. The basic concept behind EMG-driven models consists of collecting EMG data, which are filtered, rectified and input into a calibrated musculoskeletal model to predict muscle forces. However, extensive calibration trials, including optical motion capture and ground reaction force measurements, are required to define subject-specific model parameters before accurate predictions of muscle forces for individual subjects across a number of different tasks are possible. Calibration trials allow the adjustment of model parameters by minimizing the differences between joint kinematics and/or joint torques from inverse dynamics analysis and corresponding results from the EMG-driven model. In contrast to quasi-static inverse-dynamics optimisation techniques, EMG-driven models account for the dynamic force generating properties of muscles, and upon successful calibration, have been adopted to predict muscle forces during dynamic exercises such as side-stepping or dynamic lunge activities [59, 60]. Challenges remain in the placement of the electrodes and processing of the EMG signals, as well as in the calibration of reference or generic musculoskeletal models to subject- and task-specific conditions [58]. Despite these challenges, high density EMG measurements have been shown to reduce errors in muscle force predictions, but these analyses often result in an instrumentation complexity that may not be achievable during sports practice [58].

2.5 Discussion

The relevance of understanding the internal muscle forces during strength exercises becomes clear when examining the wide range of studies and research questions reported in the literature. Musculoskeletal modelling techniques have been applied to strength training to analyse the effect of altered muscle physiology on exercise performance (physiological adaptations) [62, 65, 68], the impact of exercise execution on muscle and joint forces (best choice of exercise to avoid injury)
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[15, 59, 60, 69, 70] and the internal loading state of differently sized people using the same exercise machines (safety and efficacy of equipment) [75]. Muscular weaknesses, bilateral asymmetries or changes in exercise performance have been shown to result in altered and potentially harmful internal tissue loading that cannot be investigated based on external observation or simple measurements.

Accurate assessment of the risks involved in strength exercises, and subsequent design of effective exercise schemes, are dependent upon the accurate estimation of muscle forces and joint loading during the target exercise. Different numerical techniques have been introduced to determine muscle forces in the lower extremities during strength training *in vivo*, including 1) forward dynamics analysis to study dynamic ballistic movement exercises such as the squat jump, 2) quasi-static inverse dynamics optimisation to study low-speed exercises, such as the lunge or leg press, and 3) alternative methods such as EMG-driven modelling. All methodologies are challenged by the limitations of externally measurable data and the complexity of the musculoskeletal system, i.e. the indeterminate nature of the simulation problem. Forward dynamics analyses depend upon optimisation algorithms to find the most suitable set of muscle activation levels that lead to the desired movement patterns; while muscle force calculations from inverse dynamics analyses require optimisation algorithms to distribute the net joint moments across synergistic and antagonistic muscles in a physiological manner.

The findings by studies presented in this review have provided insights into the biomechanical principles underlying strength training that would otherwise not be possible. Using musculoskeletal modelling techniques, health care related factors could be detected. For example, co-contraction forces of the hamstrings and quadriceps during squats and leg presses have been shown to significantly affect tension in the cruciate ligaments [15, 69, 70], which in turn is a crucial factor for establishing safe and effective rehabilitation programs. Furthermore, the results from musculoskeletal modelling have provided sports performance related factors, such as the muscle activation delays between stimulation onset times of proximal muscles versus plantar flexors during the squat jump and their influence on jump height deficits [66]. Lastly, the knowledge gained from computational studies has helped to support and establish training and injury prevention recommendations. For example, the reduced quadriceps muscle forces during long step lunges compared to short step lunges has supported the belief of clinicians and trainers that anterior knee translation beyond the toes during the forward lunge may be harmful to the patello-femoral joint [59].
Essentially, the forward dynamics simulation is a method of systematic trial and error, and could represent the process by which an athlete optimizes control of muscle recruitment and physiological strength for best performance of explosive movements such as the squat jump [61]. Predictive analyses based on forward dynamics provide a powerful tool to elucidate the impact of alterations in neurological activation, muscular physiology or joint alignment on performance output. As such, predictive analyses have considerable potential to improve strength training guidelines, without the need for extensive experimental measurements on individual subjects. Compared to forward dynamics simulation, quasi-static inverse dynamics optimisation techniques are computationally efficient and do not depend on EMG measurements. In particular, quasi-static optimisation techniques based on maximising muscle endurance (minimizing stress) have been widely accepted to estimate muscle forces in the lower extremities during walking and stair climbing [83, 89]. Application of the same techniques to strength training may be valid for quasi-static exercises when the training goal is maximising strength endurance. However, static optimisation techniques are generally not sufficient for predicting antagonistic muscle activity that does not occur with the goal of minimizing stress but rather to stabilize joints and maintain joint integrity [90]. EMG-driven models provide an alternative to static optimisation techniques, especially to determine muscle forces following injury or muscular fatigue where muscle recruitment patterns might be altered. However, EMG-driven models are challenged by extensive validation trials to formulate valid muscle model parameters, and difficulties remain in the placement of EMG electrodes, signal normalisation and filter choice.

Unfortunately, it remains challenging to confirm the validity of musculoskeletal models for accurately reproducing muscle forces due to the invasive nature of internal measurement techniques. Other than e.g. tendon-force measurements during surgery [83] or telemetric implants [89, 91], a gold standard for model validation remains lacking. To address this issue, an international consortium of biomechanics researchers has received funding from the National Institutes of Health to organise a series of five "Grand Challenge Competitions to Predict in vivo Knee Loads" [92]. The goal is for competitors to predict the in vivo medial and lateral knee contact forces for specific movement trials collected from subjects implanted with a force-measuring tibial prosthesis. Muscle forces are the primary determinants of joint contact forces, and thus instrumented implant data provide a direct validation of joint contact forces and an indirect validation of muscle forces. The validation of musculoskeletal models by means of instrumented joint implants has proven invaluable in orthopaedic research, for example as a basis for standardizing pre-clinical testing [89]; however, instrumented implants have exclusively been used in elderly subjects to
analyse joint loading during daily activities such as walking or stair-climbing, and have therefore not been applied to training and sports problems that involve dynamic ballistic movement loading conditions and/or impact.

It is important to note that the required degree of model complexity is dependent on the particular research question. Simplifications regarding motion, anatomy and physiology are often required in order to reduce the computational costs. However, simplified musculoskeletal models with generic material properties may lead to invalid results for certain cases. For example, the knee abduction angle during jump landing tasks was shown to be a predictor of anterior cruciate ligament injury risk in female athletes [93], whereby all rotational degrees of freedom should be involved in the knee joint during rehabilitation from anterior cruciate ligament injury; or the relationship between the measured EMG signal and the actual muscle force of the triceps surae was shown to be different for the eccentric versus the concentric contraction phase of one-legged full weight bearing ankle plantar and dorsiflexion exercises [46]. These studies suggest that all rotational degrees of freedom in the joints, or contraction-specific EMG-to-force relationships, may be required to improve modelling results in specific cases. Ideally, the influence of model complexity on the simulation results is assessed prior to making any conclusions.

Limitations of musculoskeletal simulations remain in the accurate capturing of subject-specific anatomy (e.g. segment lengths, degrees of freedom, muscle paths) and physiology (e.g. Hill-type muscle, force-length and force-velocity relationships). As a result, anatomical and physiological parameters such as the PCSA of muscles have mainly been adopted from cadaveric measurements on a few elderly human subjects (5 cadaver specimens, mean age 79.2 years in Herzog and Read [94], or 2 male cadaver specimens, mean age 82 years in Spoor and co-workers [95]), and scaled to subject-specific dimensions based on a few measurements. Lloyd and Besier [73] introduced more extensive techniques to calibrate an EMG-driven model to subject-specific conditions; however, the physiological basis of the calibration process was questioned and seemed rather tedious for application to sports practice. In the broader field of human motion analysis, increased efforts have been directed towards developing efficient computational techniques to create subject-specific anatomical models based on magnetic resonance images [96, 97]. The future application of such techniques to strength training could certainly provide a basis for analysing the influence of individual differences in muscle physiology and anatomy on exercise performance. Furthermore, subject-specific customisation of muscle model properties may be based on supplementary parameters from ultrasound or dynamometry. Such parameters could include the type of muscle fibre, maximal isometric or dynamic muscle force or physiological
performance parameters such as maximal power.

Substantial efforts have been directed towards translating musculoskeletal modelling techniques into user-friendly tools to facilitate their application in clinical and sports practice. In particular, the open source software OpenSim, developed and maintained on https://simtk.org/ by the NIH National Center for Physics-Based Simulation of Biological Structures (Simbios, Stanford University, California, USA), has significantly contributed to the uptake of computational biomechanics by the non-expert [49]. OpenSim provides a user-friendly interface for coupling forward dynamics, quasi-static inverse dynamics, and EMG-driven modelling with subject-specific experimental data to calculate joint and muscle dynamics during human movement. The development of easy-to-use and freely available software such as OpenSim marks a significant step towards the more wide-spread application of advanced computational techniques for improving the efficacy and safety of strength training. Interestingly, only a few studies have reported actual muscle force values, even though many have outlined computational techniques that entailed the calculation of muscle forces (Table 2.2). It appears that muscle forces are often used only as intermediate parameters to analyse e.g. maximal jump height through forward dynamics simulation; or quantify joint forces and moments through inverse dynamics analyses. As a result, it seems that the potential of muscle force calculations to provide evidence for the efficacy and safety of strength training programmes has not yet been fully grasped, possibly due to the difficulties in validating the presented solutions, thus explaining the limited confidence for uptake.

Knowledge of the acting muscle forces during strength training based on musculoskeletal modelling offers a means to establish effective and safe training guidelines to achieve specific training aims such as improved inter-muscular coordination, monitoring muscular changes, eliminating or preventing unbalanced muscular adaptation, point out injury predictors, as well as improve efficiency and safety of exercise equipment. In the future, it might be possible to model subject-specific loading conditions during exercising and further use the predicted muscle forces as boundary conditions for finite element continuum models in order to estimate the biological tissue adaptation due to training. The application of existing musculoskeletal modelling techniques to strength training of the lower limbs is particularly attractive because of standardised conditions and simple movement patterns that are often associated with strength exercises. A key factor in limiting the applicability and uptake of the musculoskeletal modelling techniques to clinical and sports practice seems to be the lack of experimental solutions for validating internal forces, stresses and strains in vivo. Cross-institutional initiatives, like the grand challenge to determine in vivo knee loads, should be extended to include more subjects in different age
groups and a range of different activities including strength exercises. Upon successful model validation, the quantification of muscle forces during strength training based on inverse and forward dynamic analyses are expected to help improve current training guidelines to the benefit of coaches, clinicians, athletes and patients alike.
Declarations

Ethics approval and consent to participate

All authors abide by the Ethics Committee of ETH Zurich, Switzerland ethical rules of disclosure.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this paper.

Competing interests

The authors declare that they have no competing interests.

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Authors’ contributions

All authors contributed to conception and design, paper preparation, read, and approved the final paper.
Chapter 3

Kinetic and Kinematic Differences between Deadlifts and Goodmornings

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3.1 Abstract

3.1.1 Background

In order to improve training performance, as well as avoid overloading during prevention and rehabilitation exercises in patients, the aim of this study was to understand the biomechanical differences in the knee, hip and the back between the exercises "Goodmornings" (GMs) and "Deadlifts" (DLs).

3.1.2 Methods

The kinetics and kinematics of 13 subjects, performing GMs and DLs with an additional 25% (GMs), 25% and 50% (DLs) body weight (BW) on the barbell were analysed. Using the kinetic and kinematic data captured using a 3D motion analysis and force plates, an inverse approach with a quasi-static solution was used to calculate the sagittal moments and angles in the knee, hip and the trunk. The maximum moments and joint angles were statistically tested using ANOVA with a Bonferroni adjustment.

3.1.3 Results

The observed maximal flexion angle of the knee was $5.3 \pm 6.7^\circ$ for GMs and $107.8 \pm 22.4^\circ$ and $103.4 \pm 22.6^\circ$ for DLs with 25% and 50% BW respectively. Of the hip, the maximal flexion angle was 25% smaller during GMs compared to DLs. No difference in kinematics of the trunk between the two exercises was observed. For DLs, the resulting sagittal moment in the knee was an external flexion moment, whereas during GMs an external extension moment was present. Importantly, no larger sagittal knee joint moments were observed when using a heavier weight on the barbell during DLs, but higher sagittal moments were found at the hip and L4/L5. Compared to GMs, DLs produced a lower sagittal moment at the hip using 25% BW while generating the same sagittal moment at L4/L5.

3.1.4 Conclusion

The two exercises exhibited different motion patterns for the lower extremities but not for the trunk. To strengthen the hip while including a large range of motion, DLs using 50% BW should
be chosen. Due to their ability to avoid knee flexion or a knee flexion moment, GMs should be preferentially chosen over DLs as ACL rupture prevention exercises. Here, in order to shift the hamstring to quadriceps ratio towards the hamstrings, GMs should be favoured ahead of DLs using 50\% BW before DLs using 25\% BW.

**Keywords**

Trunk movement; Spine; Motion analysis; Loading conditions; Free weights
Chapter 3: Biomechanical Differences between Deadlifts and Goodmornings

3.2 Background

Strength training exercises such as Deadlifts (DLs) or Goodmornings (GMs) are commonly used in prevention programs for reducing the risk of ACL injury or for rehabilitating low back pain patients, as well as during training to increase an athlete’s specific performance, where the loading conditions play an important role on both the passive and active musculoskeletal structures. Here, exercise kinematics play a key role for governing the lifting mechanics, and therefore modulating the risk of injury and level of performance [98]. In 1999 in Switzerland, the most frequent injuries during fitness training were the shoulder 24.4%, back 16.6%, thigh 11.0% and knee 8.8% [14]. The reasons for injury were predominantly attributed to overloading (45.6%) and wrong execution of the exercise (21.1%) [14]. Despite these statistics, a complete biomechanical understanding of the loading conditions of many exercises during strength training remains lacking.

The DL is a multi-joint resistance exercise that is performed in a variety of training settings [99]. It begins with the lifter in a squat position, with arms straight and pointing downwards, with an alternating hand grip on the bar [98]. The movement includes mainly an extension of the knee and hip until the body reaches an upright standing position. The lifting exercise then uses the following muscles: *gluteus maximus*, *erector spinae*, hamstrings, quadriceps, *trapezius*, *rhomboideus*, *deltoides* and finger flexors [33]. Due to the fact that the DL is a closed chain exercise [100], it is often used in the prevention of and rehabilitation after anterior cruciate ligament (ACL) reconstruction to improve strength of the muscular structures that surround the knee and hence dynamic stability of the joint [101–103]. The DL is also one of the three disciplines in power-lifting. The biomechanics of the lift have been studied extensively during competition, focusing on the sumo and conventional styles [98, 101, 104], where the maximal isometric forces in four different positions during DLs were shown to result in a higher potential to increase the force toward the end of the DL (from 3380 to 5829N) [105]. Training using these exercises has also been clearly related to functional adaptation of the spine, where the annual lifted loads of power-lifters has been shown to correlate with the bone mineral content in L3 [106]. However, the increased forward trunk tilt during DL lift-off may predispose the spine and back musculature to an increased risk of injury [106, 107]. In response to this, Cholewicki and co-workers [107] demonstrated that a more upright trunk at lift-off is able to reduce anterior shear force at the lumbar L4/L5 joint. Furthermore, Escamilla and co-workers [98] showed the importance of keeping the barbell mass as close as possible to the body in order to minimise
injury risk to the back as well as to enhance performance. No statistically significant differences were found in this study between the kinematics of high and low-skilled lifters, but they did show differences regarding how the barbell passed the knee: Highly skilled lifters kept the barbell mass closer to the body than less-skilled lifters.

The GM exercise is an assistance movement utilized primarily by weight lifters to strengthen the extensors of the torso, the *gluteus*, hamstrings and *erector spinae* [108]. Starting in an upright standing position and with the barbell on the shoulders, the hips are progressively flexed until maximum hip flexion is reached, but the knees remained straight throughout. GMs are a good exercise for specifically conditioning lumbar-thoracic flexion and extension of the back [109], but for all level of performance, good lifting technique is required when approaching near maximal effort to avoid acute injury or long-term damage. Here, the low back must have sufficient strength to keep the body in the correct position, since high *erector spinae* forces are known to occur, resulting in high shear and compressive forces at the level of L5/S1 [110]. The authors stated the importance of sufficiently conditioned lower back musculature and proper sport technique for reducing the risk of back injury [110, 111]. In the review of Carpenter and Nelson [112], the recommendation for low back pain patients was to train using lumbar extension reconditioning exercises with the pelvis stabilized in a specific, progressive and intensive manner, since this was shown to lead to the most favourable improvements in low back strength, muscle cross-sectional area and vertebral bone mineral density. The latter recommendation is in agreement with the general finding that strength training is able to relieve low back pain [113].

During lifting, it is well known that the main part of the axial loading of the spine is due to the large muscle and ligament forces applied over small internal lever arms. Importantly, it is thought that the amount of lumbar flexion (reduction in lordosis) determines the amount of ligamentous involvement in internal loading generation [114, 115], which may or may not be present during heavy lifts [116]. Due to the smaller lever arms of the ligaments compared to the muscles, preserving sufficient lordosis when lifting can reduce the bone-on-bone loading between the vertebral bodies due to lower posterior ligament tension [116]. However, the preservation of 1-3° margin from full lumbar flexion seems to be sufficient to avoid overloading, and this is consistent with the kinematics observed in highly skilled lifters [116].

GMs and DLs are comparable in their ability to train agility, speed and power in all sport types [117], including typical strength exercises for ACL rehabilitation [118], but also for potential injury risk during exercising [119]. Despite the widespread use of GMs and DLs, the critical
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differences in lower limb and trunk motion, and more importantly the resulting loading conditions on the joints, during GMs and DLs remain unknown. This study therefore aimed to compare the segment kinematics and joint moments of the lower limbs and spine during the entire lifting action and at the point of deepest flexion during GMs and DLs in the sagittal plane.

3.3 Methods

Nine male and four female subjects with experience in weight training (average age 24.5 ± 4.3 years, mass 74 ± 11 kg, height 180 ± 7 cm) were analysed while performing DLs and GMs exercises. The study was approved by the Ethics committee of ETH Zurich, Switzerland (EK 2012-N-57). One subject provided written informed consent to the publication of their images and all subjects provided written informed consent to participate in the study.

To analyse the motion of the body, an opto-electronic system (Vicon, Oxford Metrics Group, UK) with twelve cameras (MX40) and a sampling frequency of 100 Hz was used. The ground reaction forces were measured using two 400x600 mm force plates (type 9281B Kistler, Winterthur Switzerland), one under each foot, with a frequency of 2 kHz. The IfB Marker Set [86], consisting of 55 markers on the legs, pelvis, shoulder and arms, 22 on the back and 2 attached to the barbell, was used (Figure 3.1). The markers near the spine and on the rear and forefoot had a diameter of 9 mm while 14 mm markers were used on all other segments. Each marker was attached after palpation using double sided skin friendly tape by trained personnel.

Subjects wore their normal shoes for fitness training and shorts, while females additionally wore a bikini top. After a gentle warm-up session running on a treadmill or lifting an unloaded bar, the subjects’ performed the basic motion tasks [86]. The subject received standardized instructions for the two exercises (Table 3.1). Both exercises started in the upright position. For the GMs, the subjects were advised not to bend their knees. A set of GMs with eight repetitions and an extra load of 25% body weight (BW) on the bar was then performed with the bar positioned on the upper trapezius muscle. Afterwards, the subjects additionally performed sets of eight repetitions each for DLs with 25% and 50% BW on the bar, with the extra loads representing a typical loading of a healthy non-powerlifters and was normalised to percentage of BW.

The motion data were reconstructed in Vicon Nexus (version 1.7.1, Oxford Metrics Group, UK). The definition of a repetition of both of the two exercises was based on the start and stop point by using the vertical velocity of the two markers attached on the barbell ($v_{\text{barbell}} > 0.04 \text{ m/s}$).
Figure 3.1: **Measurement setup including the following:** a) subject with the IfB Marker Set, b) barbell, c) force plates under each foot and d) 1 of the 12 Vicon cameras.
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Table 3.1: Standardised instructions for DLs and GMs

<table>
<thead>
<tr>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>General instructions</td>
</tr>
<tr>
<td>1 Stand upright with your feet approximately shoulder width apart.</td>
</tr>
<tr>
<td>2 Point the feet slightly outward, following the natural divergence of the feet.</td>
</tr>
<tr>
<td>3 Lift the thorax to a natural spine position.</td>
</tr>
<tr>
<td>4 Hold tension in the core muscles during execution of the exercises.</td>
</tr>
<tr>
<td>5 Breathe out during the ascent.</td>
</tr>
<tr>
<td>6 Perform the exercise at the same normal speed during the downward and upward movements.</td>
</tr>
<tr>
<td>7 Lowest point before turn: No flexion in the lumbar spine.</td>
</tr>
<tr>
<td>DL specific instructions</td>
</tr>
<tr>
<td>1 Hold the barbell with a comfortable grip, one hand in a supinated and the other in a pronated position.</td>
</tr>
<tr>
<td>2 Keep the head in a horizontal view.</td>
</tr>
<tr>
<td>GM specific instructions</td>
</tr>
<tr>
<td>1 Put the barbell on the rear <em>musculus deltoideus</em> and hold it in a comfortable hand position.</td>
</tr>
<tr>
<td>2 Keep the head in extension of the spine.</td>
</tr>
</tbody>
</table>

The repetitions were time normalized and averaged. In addition, the maximum value for each repetition was averaged. Additionally, the force plates were specifically calibrated to allow for correction of the centre of pressure (COP) [120] and hence maintain accuracy during the inverse approach. The joint centres of the knee and hip were functionally determined from the basic motion tasks [86], and the joint centre of L4/L5 was defined anatomically based on anthropometric data [121]. The external joint moments in the sagittal plane were calculated using an inverse approach with a quasi-static solution [122], taking the ground reaction force and kinematic data into account [123], and normalized to BW [84]. The flexion / extension moments at the knees and hips were averaged over both limbs. The inverse approach included the position of the joints, the forces acting on each foot, and the gravitational force of the segments [84]. Due to slow accelerations of the segments during these exercises, the inertia forces were neglected. All calculations were performed in Matlab (version 8, The MathWorks Inc., Natick, MA, USA).

The position and orientation of each segment was determined relative to the reference segments defined by the standing trial as the neutral position (0° rotation) using a least-squares fit of the corresponding marker point clouds [86]. Joint rotations were described using a helical axis approach and for clinically interpretable rotational components, the attitude vectors were decomposed along the axes of a segment fixed, orthogonal, anatomically defined joint coordinate system [86]. The joint angles, the curvature and the moments were all analysed in the sagittal...
plane.

The influences of the extra barbell load (25% BW and 50% BW) and the type of lift (DLs and GMs) on the maximal segmental angles of the knee and hip, on the maximal and minimal segmental angle of the pelvis relative to lumbar spine and lumbar spine relative thoracic region, on the corresponding ranges of motion (RoMs) of the knee, hip, lumbar and thoracic spines (segmental and curvature approach) as well as on the normalized maximal moments of the knee, hip and L4/L5 were analysed using a multiple repeated-measures ANOVA (with significance defined at $p<0.05$). Bonferroni adjustment, for the three groups, as well as for the number of parameters, was then conducted to establish significant differences, resulting in significance defined at $p<0.0033$. All eight repetitions of every subject were averaged for the statistical analyses. Statistical calculations were performed using IBM SPSS software (version 21, SPSS AG, Zurich, Switzerland).

### 3.4 Results and Discussion

#### 3.4.1 Knee and Hip

**Kinematics**

The observed maximal knee and hip rotations in the sagittal plane, as well as their RoMs, were significantly smaller during GMs than during DLs (Table 3.2). The typical observed intrasubject standard deviation over eight repetitions of the maximal joint angles of the knee and hip as well as of their RoM was $<2.5^\circ$. These results in the knee were expected due to the type of lifting execution during GMs, where the knee remains almost straight. The smaller flexion movement of the hip during GMs could be a restriction due to the limited length of the two-joint hamstring muscles in the extended knee position. The obtained maximal knee angles during DL were slightly larger than those observed by Brown and Abani [124], while the hip angles remained comparable. No changes in the maximal knee and hip angles or their corresponding RoMs were found between the loading conditions with 25% and 50% BW during DLs. The observed RoMs during DLs of the knee and hip were in agreement with those observed in the study by McGuigan and Wilson [104].
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Table 3.2: Maximal segmental flexion angle of this and other studies \[98, 101, 104, 124\], range of motion (RoM) of knee, hip, pelvic-lumbar and lumbar-thoracic rotations [\[°\]] in the sagittal plane as well as the RoM of the curvatures [1/m] of the lumbar and thoracic spine

<table>
<thead>
<tr>
<th>Study</th>
<th>GM (25%)</th>
<th>DL (25%)</th>
<th>DL (50%)</th>
<th>DL (269%)</th>
<th>DL (197%)</th>
<th>DL (181%)</th>
<th>DL (289%)</th>
<th>DL (254%)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Maximal segmental knee [°]</strong></td>
<td>5.3 ± 6.7</td>
<td>107.8 ± 22.4</td>
<td>103.4 ± 22.6</td>
<td>57.5 ± 5.7</td>
<td>69.2 ± 8.5</td>
<td>59.0 ± 12.0</td>
<td>56.0 ± 9.0</td>
<td>60.0 ± 10.0</td>
</tr>
<tr>
<td><strong>Maximal segmental hip [°]</strong></td>
<td>75.3 ± 9.2</td>
<td>103.2 ± 4.2</td>
<td>101.7 ± 6.0</td>
<td>110.9 ± 5.1</td>
<td>116.3 ± 5.0</td>
<td>124.0 ± 8.0</td>
<td>108.0 ± 21.0</td>
<td>113.0 ± 5.0</td>
</tr>
<tr>
<td><strong>Flexion angle pelvis-lumbar [°]</strong></td>
<td>23.8 ± 6.3</td>
<td>27.4 ± 5.4</td>
<td>25.7 ± 4.7</td>
<td>6.8 ± 2.2</td>
<td>6.1 ± 3.1</td>
<td>6.8 ± 2.3</td>
<td>5.9 ± 3.1</td>
<td>6.0 ± 3.1</td>
</tr>
<tr>
<td><strong>Flexion angle lumbar-thoracic [°]</strong></td>
<td>-8.1 ± 7.5</td>
<td>-4.7 ± 7.3</td>
<td>-3.2 ± 8.2</td>
<td>-7.1 ± 3.7</td>
<td>-7.2 ± 3.8</td>
<td>-5.0 ± 8.0</td>
<td>-5.8 ± 10.9</td>
<td>-6.0 ± 10.9</td>
</tr>
</tbody>
</table>

*: significantly different from GM

---

Table 3.2: Maximal segmental flexion angle of this and other studies [\[°\]] in the sagittal plane as well as the RoM of the curvatures [1/m] of the lumbar and thoracic spine.
Chapter 3: Biomechanical Differences between Deadlifts and Goodmornings

Kinetics

During DLs, no changes between the two loading conditions (25% and 50% BW) on the barbell were found in the maximum moment about the knee in the sagittal plane. This seems rather surprising; one might expect higher loading due to the additional load on the bar (Table 3.3). However, it seems that a slight change of the trunk position is able to considerably modify the sagittal moments about the knee and therefore negate the effect of the extra load. In doing so, the subjects have managed to avoid additional loading at the knees. On the other hand, the hip flexion moment, as expected, increased significantly with additional weight on the barbell during DLs (Table 3.3).

During GMs, an external extension moment acted at the knee, while during DLs, from a knee flexion angle of 25° and higher, the moment produced was a flexion moment (Figure 3.2). The maximum external knee moments during DLs (Table 3.3) were comparable to knee moments in the studies of Escamilla and co-workers [98, 101], although they used much higher barbell loads. Contrary to this finding, the studies of Cholewicki and co-workers [107] and Brown and Abani [124] showed slightly lower knee moments in comparison to the present study (Table 3.3).

With the same extra load (25% BW), the sagittal moments about the hip were significantly larger during GMs compared to DLs (Table 3.3). However, the sagittal hip moments calculated for the current study were 2 to 6 times smaller than the aforementioned studies [98, 101, 107, 124] (Table 3.3), but this is in line with the reduced barbell loading and was therefore entirely expected.

The largest extension moment at the knee was, in fact, observed during GMs (Table 3.3). It should be noted that GMs are a rather isometric exercise for the knee flexors, but the hamstrings undergo eccentric and concentric contraction due to motion at the hip (Figure 3.2). At the same extra weight, the RoM of the hip throughout the exercise was significantly smaller but the hip sagittal moment significantly larger during GMs compared to DLs (Figure 3.3). The largest RoMs and the highest sagittal moment in the hip were observed during DLs with 50% extra load (Figure 3.3).
Table 3.3: Mean normalized moments and standard deviations (SD) [Nm/kg] in the sagittal plane about the knee, hip and L4/L5 region for the GM with 25% extra load, DL with 25% and 50% extra load, respectively and corresponding results from other studies [98, 101, 107, 110, 124].

<table>
<thead>
<tr>
<th></th>
<th>Knee</th>
<th>Hip</th>
<th>L4/L5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Burnett, 2002</td>
<td>-0.96 ± 0.21</td>
<td>1.63 ± 0.14</td>
<td>2.75 ± 0.26</td>
</tr>
<tr>
<td>Escamilla, 2001</td>
<td>1.11 ± 0.39*</td>
<td>1.40 ± 0.13*</td>
<td>2.81 ± 0.27</td>
</tr>
<tr>
<td>Escamilla, 2000</td>
<td>1.14 ± 0.45*</td>
<td>1.92 ± 0.19*</td>
<td>3.78 ± 0.97</td>
</tr>
<tr>
<td>Cholewicki, 1991</td>
<td>4.86 ± 1.79</td>
<td>7.80 ± 1.79</td>
<td>3.78 ± 0.72</td>
</tr>
<tr>
<td>Brown, 1985, $</td>
<td>0.78 ± 0.03</td>
<td>2.19 ± 1.91</td>
<td>3.77 ± 0.43*</td>
</tr>
<tr>
<td>Brown, 1985, $</td>
<td>0.72 ± 1.02</td>
<td>- 0.11 ± 0.08</td>
<td>- 0.72 ± 1.02</td>
</tr>
<tr>
<td>RDL: Romanian deadlift</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

*: significantly different from GM (25%)
+: DL (50%) significantly different from DL (25%)

GM: Good morning
DL: Deadlift
RDL: Romanian deadlift
LO: Lift off

This Study

<table>
<thead>
<tr>
<th></th>
<th>Knee</th>
<th>Hip</th>
<th>L4/L5</th>
</tr>
</thead>
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<tr>
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<td>-0.96 ± 0.21</td>
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</tr>
<tr>
<td>RDL: Romanian deadlift</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

*: significantly different from GM (25%)
+: DL (50%) significantly different from DL (25%)

GM: Good morning
DL: Deadlift
RDL: Romanian deadlift
LO: Lift off

$ normalized to body weight and directed by $ for the knee and hip

GM with 25% extra load, DL with 25% and 50% extra load, respectively and corresponding results from other studies [98, 101, 107, 110, 124].
Figure 3.2: Normalized knee moments in the sagittal plane [Nm/kg] (positive for external knee flexion moment) averaged over all repetitions for all subjects during Goodmornings and Deadlifts with corresponding knee flexion angle (°), zero represents a straight leg, defined on the basis of the standing trial) compared to the squat exercise (data taken from [84]). *: Starting point of the eccentric phase at the upright position; blue: DL with 25% extra load; red: DL with 50% extra load; green: GM with 25% extra load; blue dotted: squats with 25% extra load; red dotted: squats with 50% extra load.
Figure 3.3: Normalized hip moment in the sagittal plane [Nm/kg] averaged over all repetitions and all subjects (positive for external hip flexion moment) with corresponding hip flexion angle (°), zero represents a straight hip, defined on the basis of the standing trial) compared to the squat exercise (data taken from [84]).*: starting point of the eccentric phase at the upright position; blue: DL with 25% extra load; red: DL with 50% extra load; green: GM with 25% extra load; blue dotted: squats with 25% extra load; red dotted: squats with 50% extra load.
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The motion and loading patterns of DLs were observed to be similar to squats [84]. Here, the maximum knee flexion angle was within a few percent (<4%) whereas the maximal flexion moment in the knee was higher for squats with 50% extra load (Figure 3.2). Larger differences were observed in the hip, however, where the maximal flexion moment during DLs was at least 50% larger (Figure 3.3).

**Relevant outcomes for ACL injury prevention**

A number of studies have identified the force ratio of the quadriceps to hamstring (H:Q) as a risk factor for ACL rupture [125], especially in women [126]. Holcomb and co-workers [126] used, amongst other exercises, GMs and straight leg DLs to modify the H:Q ratio by training within 6 weeks. For the training of multi-joint muscles, such as the *M. semimembranosus* and *semitendinosus*, the joint angles of the hip and the knee, as well as their corresponding moments, should be taken into account. Based on the finding that maximal external extension moment in the knee (Figure 3.2) and a flexion moment in the hip were observed during GMs (Figure 3.3), it follows that GMs might provide an effective strategy for focused strengthening of the hamstrings. This finding is in agreement with the study of Ebben and co-workers [125], who demonstrated the importance of hamstring training for the potential reduction of ACL injuries and who further recommended GM training to be included as a preventative measure.

One new and notable outcome of this study is that no larger external knee moments were observed by a larger extra load on the barbell during DLs (Table 3.3). Compared to squats [84], DLs have the advantage that the flexion moment in the knee is smaller (Figure 3.2) but the flexion moment in the hip is larger (Figure 3.3) using the same extra load. Therefore, based on the observed kinetics and kinematics of the strength exercises, the following ranking is suggested in order to shift the H:Q ratio towards H: GM, DL 50%, DL 25% and squats.

### 3.4.2 Back

**Kinematics**

Neither the RoMs of the lumbar and the thoracic curvatures, nor the maximal and minimal flexion and extension angles of the pelvic-lumbar and the lumbar-thoracic segments, were affected by the execution or the extra weight on the barbell (Table 3.2).
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Kinetics

The flexion moment in the L4/L5 region was significantly higher during DLs than GMs due to the additional load on the barbell (Table 3.3). However, the two exercises produce the same loading conditions at the L4/L5 region using the same load on the barbell (Table 3.3). Previously presented data of normalized moments in L4/L5 using 259% BW load during conventional DLs [107] were larger compared to the values of the present study (Table 3.3). However, the study from Burnett and co-workers [110] showed smaller moments at L4/L5 during Romanian deadlift exercises compared to the present study, even though they used much higher normalized weights on the barbell (133% BW extra load). During GMs, Burnett and co-workers [110] found higher moments at L4/L5 compared to the present study, which could be explained by the heavier extra loading on the barbell (72% BW extra load).

Back training

Surprisingly, the relationship between the L4/L5 moment and the lumbar curvature in the sagittal plane was different between the concentric and the eccentric phases of lifting; especially during DLs using 50% extra weight (Figure 3.4). During the eccentric phase, the lumbar back maintained its higher curvature longer compared to the concentric phase with the same sagittal moment (Figure 3.4). Due to the fact that the flexion moments at L4/L5 were similar for the two exercises with 25% BW extra load (Table 3.3) and no differences in the RoM of curvature or the segmental kinematics of the back were observed (Table 3.2), from a biomechanical point of view, the two execution types are comparable for the trunk. As a result, differences between the exercises of the kinematics and mechanics in the lower limbs should be considered more relevant.
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Figure 3.4: Normalized L4/L5 moment in the sagittal plane [Nm/kg] averaged over all repetitions and all subjects (positive for external L4/L5 flexion moment) with corresponding lumbar curvature [1/m]. *: starting point of the eccentric phase at the upright position; blue: DL with 25% extra load; red: DL with 50% extra load; green: GM with 25% extra load.
3.4.3 Practical applications

In order to optimize the training effect of the quadriceps, a large RoM [127] and external flexion moment in the knee is demanded. It follows that the DL is the preferable exercise for quadriceps training, although the extra load did not affect the sagittal moment in the knee. During squatting, similar RoMs in the knee, but higher moments in the sagittal plane and a load dependency of the moments have been observed [84].

To train the M. gluteus maximus, GMs produce a higher sagittal moment but a smaller range of motion than DLs. If a large range of motion is required, DLs are therefore considered the better choice. The small RoM in the knee suggests that GMs should be chosen before DLs at the early stage of rehabilitation for subjects with a previous knee injury. Furthermore, GMs are suited to avoid external flexion moments at the knee. The magnitude of the resulting extension moments during GM is similar to the magnitude of the flexion moment during DLs.

3.5 Conclusions

DLs and GMs show different motion and loading patterns for the lower extremities, where the knee remains almost straight during GMs, hence producing a large extension moment. The maximal knee and hip angle, as well as the RoMs of the knee and hip, are smaller during GMs than DLs. Kinematically, the DL is not generally affected by the extra weight on the barbell. The flexion moment at the knee during DLs is also not influenced by the additional 25% load, however, the sagittal moment in the hip is higher during DLs using 50% BW extra load. Based on the higher sagittal moments in the hip and the L4/L5 region with higher barbell loads, great care should be taken to ensure core stability of the trunk during lifting due to high loading of the spine, especially when training with higher extra loads. Finally, for prevention of ACL injuries, GM are recommended for training the hamstrings to quadriceps ratio.
Chapter 3: Biomechanical Differences between Deadlifts and Goodmornings

Declarations

Acknowledgements

We like to thank to Dr. Hans Gerber, Marco Hitz and Peter Schwilch for the technical support.

Competing interests

The authors declare that they have no competing interests.

Authors’ contributions

FS and JL acquired the data, and undertook the data analyses including the preparation of the tables and figures. FS furthermore performed the statistical analysis and helped drafting the manuscript. RL supported the measurement set up, data analysis and helped drafting the manuscript. BT helped in both interpreting the data and drafting the manuscript. SL made the concept and design of this study, supervised the data analyses and interpretation and helped drafting the manuscript. All authors read and approved the final manuscript.
Chapter 4

Loading conditions during different executions of split squats and back extension exercises

These studies do not belong to the core competences of this thesis. However, to fulfil all examined exercises as well as to origin of the data used for musculoskeletal simulation, short abstracts of the two studies are provided here. Corresponding tables and figures are attached in the appendix.
Chapter 4: Loading conditions during split squats and back extension exercises

Joint Angles of the Ankle, Knee and Hip and Loading Conditions during Split Squats

adapted from:

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Journal of Applied Biomechanics, 30(3):373-80, 2014
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4.1 Abstract Split Squat Exercises

The aim of this study was to quantify how step length and the front tibia angle influence joint angles and loading conditions during the split squat exercise. Eleven subjects performed split squats with an additional load of 25% body weight applied using a barbell. Each subject’s movements were recorded using a motion capture system, and the ground reaction force was measured under each foot. The joint angles and loading conditions were calculated using a cluster-based kinematic approach and inverse dynamics modelling respectively. Increases in the tibia angle resulted in a smaller ranges of motion (ROM) of the front knee and a larger ROM of the rear knee and hip. The external flexion moment in the front knee/hip and the external extension moment in the rear hip, decreased as the tibia angle increased. The flexion moment in the rear knee increased as the tibia angle increased. The load distribution between the legs changed <25% when split squat execution was varied. Our results describing the changes in joint angles and the resulting differences in the moments of the knee and hip will allow coaches and therapists to adapt the split squat exercise to the individual motion and load demands of athletes. Results are presented in the figures and tables attached in the Appendix.

Keywords

Inverse dynamics, movement analysis, strength exercise
Chapter 4: Loading conditions during split squats and back extension exercises

Declarations

Ethics approval and consent to participate

Ethics approval and consent to participate. The study was approved by the ethics committee of ETH Zurich, Switzerland.

Conflict of Interest Disclosure

There is no conflict of interests.

Funding

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Acknowledgements

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Loading conditions in the spine, hip and knee during different executions of back extension exercises

adapted from:

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Institute for Biomechanics, ETH Zurich, Switzerland
Scandinavian Journal of Medicine & Science in Sports 2016
In review
4.2 Abstract Back Extension Exercises

Introduction

The back extension (BE) is a strength exercise for training the dorsal trunk and hip muscles. In order to optimise training recommendations that avoid overloading and possible injury, the aim of this study was to determine the loading conditions and the influence of different execution forms of BE on spine, hip and knee ranges of motion (RoMs), joint moments and muscle activity.

Methods

Kinematics, kinetics and muscle activity of two execution types (B Eh: dynamic hip, BEs: dynamic spine) and three versions (one-legged, two-legged, with inverse breathing pattern) of BE were measured with 3D motion analysis, force plates and electromyography (EMG) in 16 subjects. RoMs and external joint moments were calculated by means of an inverse dynamics approach and analysed with a linear mixed model.

Results

Although lumbar spine flexion was observed in both execution types, thoracic spine flexion predominantly occurred during BEs while thoracic spine extension was observed during B Eh. Larger maximal back and hip moments were observed for B Eh than BEs. The activity of dorsal back and hip muscles, as observed using EMG, was increased for one-legged executions. An inverse breathing pattern did not seem to influence the kinematics, the joint moments or EMG muscle activity. Results are presented in the figures attached in the Appendix.

Conclusion

In order to strengthen the hips and lower back, B Eh seem to be more efficient due to the higher moments, with higher or similar RoMs in the hip and lower back. One-legged BEs seem to provide an effective training for the hamstrings and hip regions without subjecting the spine to excessive loading, possibly promoting this as an effective training during training and rehabilitation.

Keywords

Strength Training, External Joint Moments, Ranges of Motion, EMG, Trunk, Rehabilitation
Declarations

Ethics approval and consent to participate

Ethics approval and consent to participate. The study was approved by the ethics committee of ETH Zurich, Switzerland (EK 2014-N-31).

Availability of data and material

The dataset generated and analysed during the current study are available from the corresponding author on reasonable request. Raw data and analysed results are stored for the next then years at the laboratory for movement biomechanics, ETH Zurich, Switzerland.

Competing interests

The authors declare that they have no competing interests.

Funding

No funding of third parties was contributed.

Authors’ contributions

FS, RH and NH acquired the data, and undertook the data analyses including the preparation of the tables and figures. NS and FS performed the statistical analysis and helped drafting the manuscript. BT helped in both interpreting the data and drafting the manuscript. FS and SL made the concept and design of this study, supervised the data analyses and interpretation and helped drafting the manuscript. All authors read and approved the final manuscript.

Acknowledgements

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Chapter 5

Robustness of kinematic weighting and scaling concepts for musculoskeletal simulation

adapted from:

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\textit{Computer methods in biomechanics and biomedical engineering} 2016

In review
5.1 Abstract

Musculoskeletal modelling is widely used to estimate internal loading conditions. In order to optimise robustness and reduce errors between the reference motion data (RMD) and the musculoskeletal simulation, 90 permutations of kinetic and kinematic data were analysed during split squats.

A ranking for the scaling and kinematic weighting concepts based on the RMS errors when including functional centres of rotation (fCoRs), joint angles, and skin markers, revealed that analyses should include fCoRs for optimal registration of the musculoskeletal model to the RMD. This investigation provides guidelines for registering musculoskeletal models to reference kinematic and kinetic data for the individualisation of simulations.

Keywords

subject specific musculoskeletal modelling; robustness; error estimation; strength training; split squats
5.2 Introduction

Musculoskeletal (MS) simulation plays a key role in biomechanics for estimating internal loading conditions, including muscle and joint contact forces. More specifically, knowledge of the internal forces has been extensively used to provide improved understanding of clinical treatments e.g. joint replacement [50, 51] and its outcome [52], or muscle replacement in crouch gait children [49], but also to estimate the loading conditions in sports and activities of daily living [48, 57, 61, 62, 68, 128]. Direct measurement of muscle and joint contact forces is currently not possible, resulting in the recent development of MS modelling techniques that are able to provide access to these parameters, albeit indirectly, by means of numerical optimization processes [129]. As a result of the detailed modelling approaches required to accurately determine the kinetics of the human body, MS simulation software packages such as Anybody (Anybody Technology, Aalborg, Denmark), OpenSim (Simtk, Standford, CA, United States; [49], Biomechanics of Body (BoB;[47]) and others have become widely available.

A common approach for generating MS simulations is to first scale a reference MS model that possesses generic anthropometrical parameters, such as height, body weight, segment lengths, muscle paths etc. to the specific subject in question. The second step is to calculate the body/segment kinematics from e.g. skin marker trajectories or joint angles captured during the target movement. Finally, the scaled model and the calculated segment kinematics are combined with measured kinetic data to allow estimation of the internal loading conditions by means of inverse dynamics and optimization processes to solve the muscle distribution problem [129].

The accurate measurement of segment kinematics is challenging due to the complexity of the soft tissues moving relative to the underlying skeletal structures [130, 131]. As a result, motion capture varies from techniques using data extracted from simple videos, through retro-reflective markers attached to the subject’s skin, to extensive measurement using e.g. video-fluoroscopy [132–134], MRI [135] or ultrasound [136]. However, due to their increasing availability, non-invasive, high-speed and accurate nature, optoelectronic infra-red measurement devices have now become a standard technique for the capture of human movement. In order to improve the robustness of MS simulations [137, 138], specific approaches for the reduction of soft tissue artefact [130] and the assessment of the underlying skeletal kinematics [139–142] have been developed [143] and are now even integrated within commercial motion capture software (e.g. Vicon).
In order to understand the subject specific loading conditions based on the measured kinematic data, approaches to register reference MS models to the individual’s anatomy and kinematics are required. However, the accuracy of a simulation is known to be sensitive to the subject specific anatomical specification, including bone and muscle architecture [135]. Therefore, the scaling processes, as well as the method of including kinematic data into the simulation, will influence the resulting internal loading conditions in a complex manner [144, 145]. With the aim of enhancing the physiological validity and accuracy of the segment kinematics, different methodologies for weighting the motion capture data [146] and/or geometrical parameters, such as human shape, bone structure or muscle paths, are also thought to improve the robustness of MS models [147]. Consequently, reconstruction of a subject’s original motion and kinetics, using specific scaled MS models involves many unknowns and assumptions, making the result highly sensitive to the considerable number of settings used during the scaling and kinematics registration.

With numerous approaches for scaling and fitting model anatomic and kinematic data, it is often unclear how the settings used during scaling and kinematics registration can best be utilised to allow the motion patterns that were actually measured to be reconstructed in the MS model. Therefore, the aim of this study was to quantify the robustness of MS simulation using different scaling methods and differently weighted kinematic concepts, as well as to estimate the resulting errors in terms of kinematics and kinetics.

5.3 Methods

5.3.1 Kinematic and kinetic data

The data used in this study was captured previously in 11 subjects for the analysis of loading conditions in the lower limbs during six repetitions of 10 different types of split squats, leading to 660 cycles [87]. Split squats are a common multi-joint strength exercise to train mainly the m. gluteus maximus, m. iliopsoas, quadriceps, and hamstrings [32] in a slow, non-impact manner. Their use in this study also allowed large knee and hip ranges of motion (RoM) to be examined. The data set consisted of 3D kinematic skin marker trajectories of 55 bone and soft tissue markers, mainly attached to the lower limb ([86]; no spine) captured using an opto-electronic infrared system (100 Hz, Vicon, OMG, Oxford, UK) together with ground reaction force data (2 kHz, Kistler AG, Winterthur, CH) (Figure 5.1).
Figure 5.1: **Study design:** A) Kinematic data from skin markers and functionally defined centre of rotations (fCoRs), as well as pre-calculated joint angles from the reference movement data (RMD) and ground reaction forces (GRFs), were used as input data. B) Scaling concepts (green) leading to 2 permutations were used to register the reference model. C) Kinematic weighting concepts (3x5x3 = 45 permutations: green) were investigated for driving the scaled musculoskeletal model. D) To assess simulation robustness, the 3D location of the fCoRs, the ranges of motion (RoM) of the joint angles, as well as maximal external joint moments were calculated and used as evaluation parameters.

High: high weighting; low: low weighting; none: no weighting; M all: manually weighted inclusion of all markers; all: weighting of all joint angles in all planes; M bone: manually weighted inclusion of markers based on bone landmarks; A bone: automatic weighted inclusion of markers based on bone landmarks; A all: automatic weighted inclusion of all markers; flexion: weighting of joint angles in the sagittal plane; Sag: sagittal plane; Front: frontal plane; Trans: transverse plane.
5.3.2 Reference movement data

While the approaches for kinematic and kinetic assessment vary considerably between motion laboratories, the approach according to List and co-workers (2013) [86] was used for functionally determining the lower limb joint centre of rotations (fCoRs) and functionally defined axes of rotation in the knee, acquired during a range of basic motion tasks. Joint angles were calculated using a direct kinematics procedure after which external joint moments for each of the 660 cycles were calculated using quasi-static inverse dynamics. The original motion data published by Schütz and co-workers (2014) [87] is here named the "reference movement data" (RMD).

5.3.3 Modelling approach

Scaled generic MS models [49] were constructed to assess the resultant joint kinetics and kinematics in the ankles, knees and hips of each subject. Here, the “Gait2392_simbody” model [50, 148–150] was adapted to include 14 body segments and one segment describing the barbell (Figure 5.2). The reference model therefore comprised 30 degrees of freedom (DoF), including 3 DoFs in each knee and ankle joint, as well as a predefined flexion dependent path of the centre of rotation (CoR) at the knee introduced by Yamaguchi and Zajac (1989) [150]. 47 skin markers were attached to the reference MS model at segment locations according to List and co-workers (2013) [86] (6 markers attached on the elbow and wrist were not included in the MS model) and two on the barbell. 21 markers were palpated on bone landmarks, where 25 markers were additionally placed on lower limb segments and subsequently handled as soft tissue markers (Figure 5.2). Furthermore, six virtual markers were also included in the model at the CoR of the hip, knee and ankle joints. The chosen coordinate systems and joint angle definitions were consistent with ISB recommendations [151] based on Grood and Suntay (1983) [152].

5.3.4 Subject specific scaling concepts

Two different scaling concepts were used to register the reference MS model: The first scaling concept was based on the standard operational procedures used in OpenSim [49]. Here, the segment dimensions were determined according to the bone landmarks. The markers of the reference model were then fitted to the captured marker cloud during an upright standing trial (Figure 5.1B), that included a marker weighting of 5 to 1 for the palpated bone to soft tissue markers respectively (Figure 5.2), as well as the pre-calculated joint angles with a weighting of
Chapter 5: Robustness of musculoskeletal simulations

Figure 5.2: The adapted "Gait2392_simbody" model with 14 body and one barbell segment resulted in 30 degrees of freedom. 21 markers were palpated on bone landmarks (highlighted with a black circle), while the remaining 25 markers were classified as soft tissue markers. In addition, 6 virtual markers were included at the CoR of the hip, knee and ankle joints.
0.02. In the second scaling concept, the segment dimensions were based on the distances between the fCoRs of the hip, knee and ankle of the RMD. Keeping the same marker weightings as in the first scaling concept, the fCoRs were additionally weighted with a factor of 60 (Figure 1B). For determination of the local coordinate system at the knee, the functionally defined knee axis of rotation was included in the scaled model.

5.3.5 Kinematic approach

For each subject (n=11) and each scaling concept (n=2), segment kinematics were calculated using the inverse kinematics procedure in OpenSim after which an inverse dynamics analysis was performed to calculate the external joint moments. In order to explore the effects of different standard weighting options available within OpenSim [49], 45 “kinematic weighting concepts” were investigated for driving the inverse kinematics solution. These differed based on skin marker weightings, the in- or exclusion of fCoRs, and the in- or exclusion of pre-calculated joint angles ([86, 87]; 5.1C). The skin markers were either neglected completely (weight being 0, Figure 1C), or assigned weightings according to four different approaches: I) only the bone markers (Figure 2) were included and weighted manually with a factor of 1; II) all markers were weighted manually with a factor of 1; III) only the bone markers were included, and weighted automatically based on soft tissue artefacts (STA), and IV) all markers were included, and weighted automatically based on STA. The STA automated weighting procedure was adapted from Heller and co-workers (2011) [146] and Kratzenstein and co-workers (2012) [153] using the relative variance of distance between each skin marker and the corresponding segment centre of mass. In order that each segment was considered with equal importance, the sum of all skin marker weighting factors was defined to be 10 on each segment, independent of the number of markers attached to that segment. The fCoRs were included as additional virtual markers and either neglected (weighted with 0, Figure 1C); weighted low (for the hip (10), knee (10) and ankle (6) joints), or weighted high (100, 100 and 60 respectively) in order to simulate almost complete dependence upon the joint centres alone. Similarly, the absolute joint angles of the pelvis and the relative joint angles between the lumbar, pelvis, thigh and tibia segments using the direct kinematics approach were either neglected, weighted only in the sagittal plane (weighting 0.02), or weighted in all planes (30 DoF; weighting 0.02). This resulted in 90 possible scaling and kinematic permutations for each of the 6 repetitions of 10 exercises in each of the 11 subjects; and therefore a total of 59400 individual MS simulations.
5.3.6 Robustness

The trajectories of the joint centres and angles from the inverse kinematics, as well as the external joint moments were compared against the RMD for all 90 scaling and kinematic permutations. To assess the effect on the kinematics, two parameters were evaluated. Firstly, differences in the 3D locations (global coordinate system) of the hip, knee and ankle joints ($\Delta d^{\text{Joint}}$) between the MS simulations (MSs) and the RMD were calculated for each cycle:

$$\Delta d^{\text{Joint}} = \sum_{i=1}^{n} |\vec{r}_{\text{Joint MSs}i} - \vec{r}_{\text{Joint RMD}i}|$$  (5.1)

where $\vec{r}$ is the 3D location of the joint (hip, knee and ankle of front and rear limbs), and $n$ is the number of frames in the respective cycle.

Secondly, in a similar manner, the differences in the RoMs ($\Delta \text{RoM}^{\text{Joint},p}$) of the individual degrees of freedom ($p$: sagittal [flexion], frontal [adduction] and transversal [rotation]; according to Grood and Suntay (1983) [152]) of each joint (hip, knee and ankle) of the front and rear limbs were calculated over each cycle as follows:

$$\Delta \text{RoM}^{\text{Joint},p} = |\text{RoM}_{\text{Joint MSs}}^{\text{Joint},p} - \text{RoM}_{\text{RMD}}^{\text{Joint},p}|$$  (5.2)

Additionally, the normalised differences $\Delta \text{RoM}^{\text{Joint},p}_{\text{norm}}$ were computed by dividing $\Delta \text{RoM}^{\text{Joint},p}$ by $\Delta \text{RoM}^{\text{Joint},p}_{\text{RMD}}$ to allow a fair comparison between the different movement types.

As a kinetic evaluation parameter, reported for the hip and knee joints only (ankle joint is of low interest during this type of strength exercise and is less affected due to the low external moment), the absolute difference in the maximum external joint moment in the sagittal plane, divided by each subject’s bodyweight, was calculated as follows:

$$\Delta M^{\text{Joint}}_{\text{max}} = |M^{\text{Joint}}_{\text{MSs max}} - M^{\text{Joint}}_{\text{RMD max}}|$$  (5.3)

The mean differences, their standard deviations (SDs), as well as the root-mean-square error (RMSE) were calculated for all 90 permutations for each parameter. To quantify the robustness of each of the scaling and kinematic options (highlighted in green in Figure 5.1), the RMSEs of
the three aforementioned evaluation parameters were averaged and the SDs were calculated for all possible permutations that include that specific option. To provide a fair comparison, each RMSE was normalised (\textit{Norm RMSE}) by dividing the RMSE by its averaged RMSE from all permutations.

5.3.7 Ranking

In order to assess the relative performance of each of the 90 different simulation permutations in a fair manner, a ranking based on each parameter’s \textit{Norm RMSE} was produced. The sum of the three normalised RMSEs ($\Delta d_{\text{Joint}}$, $\Delta RoM_{\text{Joint, p}}$, $\Delta M_{\text{Joint max}}$) was calculated (sum \textit{Norm RMSE}) and ranked according to the values, where a lower sum \textit{Norm RMSE} resulted in a higher ranking, indicating that the kinematics and kinetics of the resulting MS simulation better reflected those of the RMD.

5.4 Results

All scaling and kinematic weighting concepts could be successfully simulated in OpenSim except for the 6 permutations that involved no marker weighting and no \textit{fCoR} weighting. These permutations were therefore not taken into account for further analysis.

5.4.1 Robustness

The scaling concept that included \textit{fCoRs} resulted in smaller RMSEs in $\Delta d_{\text{Joint}}$ than without (Table 5.1). Similarly, the inclusion of highly weighted \textit{fCoRs} in the kinematic weighting concepts, and omitting other input data (i.e. markers: none, angles: none), resulted in the smallest RMSEs in $\Delta d_{\text{Joint}}$. Regarding RMSEs of $\Delta RoM_{\text{Joint}}$, the inclusion of all precalculated angles in the simulations showed similar RoMs to the RMD and was at least 5 times smaller than when the pre-calculated angles were excluded. Moreover, this concept subgroup had the smallest SD (0.1°) suggesting one of the most robust combinations. On the other hand, neglecting all skin markers as kinematic input data resulted in large RMSEs for $\Delta RoM_{\text{Joint}}$ as well as for $\Delta M_{\text{Joint max}}$. Furthermore, neglecting pre-calculated joint angles resulted in large RMSEs in $\Delta M_{\text{Joint max}}$. In- or exclusion of all other parameters did not seem to play a key role for the evaluation parameters. Compared to the mean RMSEs of $\Delta M_{\text{Joint max}}$, large SDs were observed in the permutations where
fCoRs were included, but also where markers or joint angles were excluded as kinematic input data.

Table 5.1: Averaged RMSEs resulting from the scaling and weighting permutations, including the corresponding SDs. The different columns distinguish the three evaluation parameters: the 3D location of the joint centers ($\Delta d_{Joint}$), the range of motions ($\Delta RoM_{Joint}$) and maximal external joint moments ($\Delta M_{Joint,max}$). Each of these columns is separated by the two scaling permutations. The first row (All) shows the averaged difference (between the reference and simulated data) of all kinematic weighting permutations, while the following rows show grouped mean values according to the kinematic weighting concepts used.

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Joint Centres: Looking more specifically at the different joints, a large variation in the difference of 3D locations of the hip, knee and ankle joints ($\Delta d_{Joint}$) was observed (Figure 5.3). However, the average $\Delta d_{Joint}$ was similar for each the front and the rear limbs in all joints. The mean RMSE for the hip, knee and ankle joints between the MS simulation and RMD was about 13.6mm (SD ± 4.3mm).

RoMs: The means of the minimal differences in the range of motion in each joint ($\Delta RoM_{Joint,p}$) were below 0.5° in all joints and all planes, except for adduction and rotation of the rear hip and rotation of the front hip. However, more unstable permutations resulted in unrealistic angles with extreme maximum errors of up to 98° (Table 5.2). Similar to $\Delta d_{Joint}$, comparable angles were found between the joints of the front and trailing limbs. The SDs in all joints and all planes were almost as high as the mean values themselves (Table 5.2), indicating large differences between the smallest and highest values compared to the RMD.

Moments: The mean as well as the maximum $\Delta M_{Joint,max}$ were higher in the knee than in the hip.
Figure 5.3: Boxplot of the RMSE of the differences in the 3D locations of the hip, knee and ankle joints (RMSE $\Delta d_{\text{joint}}$) of the front and rear leg between the reference movement data and simulated locations for all 90 permutations.
Table 5.2: The mean values of all 90 permutations, its SD, and the RMSE of the differences in the range of motions (RoMs) of the hip, knee and ankle joints in each anatomical plane ($\Delta RoM^{Joint,p}$), as well as the normed differences ($\Delta RoM^{Joint,p}_{norm}$), are given for the front (above) and rear (below) limbs. Out of all 90 permutations, the maximal values are shown separately for all joints and planes. (flex: flexion, add: adduction, rot: rotation, inv: inversion).

<table>
<thead>
<tr>
<th>Joint,p</th>
<th>$\Delta RoM^{Joint,p}$ [°]</th>
<th>$\Delta RoM^{Joint,p}_{norm}$ [°]</th>
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Chapter 5: Robustness of musculoskeletal simulations

(Figure 5.4). RMSEs of between approximately 0.05 and 0.1 Nm/BW were observed across all four joints, thus representing 3 to 9% of the absolute joint moments. Similar to the RoM, more unstable permutations resulted in unrealistic joint loading conditions with extreme maximum errors up to 1.3 Nm/BW.

Figure 5.4: Boxplot of the RMSE of the absolute difference (between the reference and simulated data) in the maximum external joint moment in the sagittal plane, normalised to each subject’s bodyweight ($\text{RMSE} \Delta M_{\text{Joint max}}$) for all 90 permutations in the hip and knee joints of the front and rear limbs. For comparison, the blue lines report the ranges of absolute external moments over all 10 types of split squats analysed by Schütz and co-workers (2014) [87], and demonstrate that in certain cases, the error is as large as the measured values.

5.4.2 Ranking

The concepts leading to the highest 20 rankings and that therefore best fit the RMD all used the scaling concept with the inclusion of $f\text{CoRs}$ (Table 5.3). In addition, the highest 17 rankings included $f\text{CoRs}$ in the inverse kinematics procedure, where the first 9 (except the permutation
ranked 4) all used a high weighting. The inclusion or exclusion of the skin markers seemed to make little or no difference to the total error. Furthermore, the concepts leading to the first 6 rankings all used the inclusion of pre-calculated angles in all planes. Importantly the rankings 7-9 all excluded (none) the use of joint angles, but were otherwise the same permutations as 3, 5 & 6 regarding the scaling concepts and the inclusion of fCoR and skin marker weightings. These permutations (rank 7-9) produced similar levels of total error (Sum Norm RMSE between 1.36 and 1.39) compared to the permutations ranked 1-6; however, the errors (Norm RMSEs) of each parameter ($\Delta d_{\text{Joint}}$, $\Delta RoM_{\text{Joint}}$, $\Delta M_{\text{Joint}}^{\text{max}}$) were distributed differently compared to the errors seen for ranks 1-6. In fact, this trend was also more generally observed where $\Delta d_{\text{Joint}}$ and $\Delta RoM_{\text{Joint}}$ both displayed a clear dependency on the usage of joint angles. Here, $\Delta d_{\text{Joint}}$ was larger in permutations that included joint angles, while $\Delta RoM_{\text{Joint}}$ was lower, and vice versa.

5.5 Discussion

With musculoskeletal modelling simulations becoming increasingly available, access to an individual’s internal loading conditions now opens perspectives for improvements in subject specific training, rehabilitation regimes and even targeted therapies. However, with a variety of potential errors for relating the observed motion capture to the scaled generic MS model in an inverse kinematics approach, it remains unclear how best to achieve robust MS modelling analyses in terms of producing minimal errors between kinematics calculated using direct and inverse kinematics procedures. As such, we aimed to determine the parameters that allow scaled generic musculoskeletal models to be registered to the recorded motion data in the most robust manner possible, but also understand the levels of error involved. Here, open source musculoskeletal simulation software (OpenSim SimTk; [49] was used to explore the effects of 90 different permutations of scaling and kinematic weighting concepts on the robustness of the resulting MS analyses.

Our results indicate an overall difference (mean of 45 permutations) in the 3D locations of the hip, knee and ankle joints ($\Delta d_{\text{Joint}}$) throughout complete activity cycles (in this case squats) of 8.5 mm when functionally derived joint centres were included in the scaling process, whereas mean registration errors up to 18.0 mm were observed with their exclusion (Table 5.1). These results are comparable to values reported in the literature varying between 8-16 mm [154–157] for approaches that also included functional joint centres, and up to 38 mm [158] for those that did not. Similarly, in the estimation of joint centres, errors of 19-36 mm have been observed using
Table 5.3: Rankings were assigned to each permutation based on the parameters’ normed RMSEs (Norm RMSE). Each parameter was ranked according to the total error i.e. the difference in 3D location of the joint centres of rotations, angles and moments (sum norm RMSE), where a lower sum norm RMSE resulted in a higher ranking, indicating that the MS simulation was closer to the reference motion data.

<table>
<thead>
<tr>
<th>Ranking</th>
<th>Scaling concept</th>
<th>Kinematic weighting concept</th>
<th>Norm RMSE</th>
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Note: No numerical solution achieved
anthropometrical data [158], suggesting that the source of different errors is consistent with our analyses using permutations without the inclusion of fCoRs.

Joint Centres: Despite the very different kinematics of the lower limbs during the split squat activities, the average $\Delta d^{Joint}$ was similar for each joint. The lowest single RMSE of $\Delta d^{Joint}$ of all permutations was 1.8 mm, and was found in the hip joint of the front leg (Figure 5.3), values that were comparable to the errors reported in other studies, including Heller and co-workers (2011) [53], whose weighted optimal common shape technique achieved hip joint centres with a precision of $3.4 \pm 1.1$ mm. The lowest average errors over all permutations observed for a joint, however, were 10.9 mm, found in the ankle joint of the rear leg – an error value that could be considered representative of general or standard procedures that aim to reproduce measured kinematics in MS models. Since the CoR of the ankle joint has only low displacement in space, as well as the lowest soft tissue coverage compared to the hip and the knee [159–161], it is entirely reasonable that this joint experiences the lowest STA and therefore exhibited the lowest mean differences to the RMD. However, based on the presented rankings, targeted weighting and scaling of measurement input during scaling and inverse kinematics can reduce considerably the errors down to a few millimetres, or up to a factor of 8 compared to their mean RMSE values (hip of front limb).

RoMs: Large RMSEs and SDs compared to the mean $\Delta RoM^{Joint,p}$ (Table 5.2) indicate only low levels of robustness in this parameter. Here, since mean and RMSE outliers of up to 98° were found, most likely due to non-physiological or unnatural simulation postures resulting from under-determined boundary conditions, it is important to avoid unrealistic segment "flipping", and ensure appropriate RoMs for each joint within each anatomical plane. Importantly, permutations that excluded the use of skin markers and joint angles resulted in unrealistic solutions (rankings 81-84, Table 3), while permutations that excluded fCoRs and skin markers were even worse, whereupon no numerical solution could be achieved (rankings 85-90). In studies investigating the effects of soft tissue artefacts on real segment kinematics using photogrammetric and fluoroscopic approaches, comparable errors of up to 192% and 117% were found in the RoM for knee abduction–adduction and internal–external rotation respectively [162]. Similarly, the observed RMSEs for flex/ex: 7.3 - 47.3%; int/ext: 46.7 – 102.3%; abd/add: 29.3 – 93.5% for open-chain knee flexion, hip axial rotation, level walking, and step-up exercises suggest that such errors are not only joint, but also activity dependent ([163]; c.f. Table 5.2). However, the lowest $\Delta RoM^{Joint,p}$ of all 90 permutations in our study were below 0.5° in all planes, indicating a good representation of the RMD, hence signifying that robust solutions can be achieved when
permutations including pre-calculated angles in all planes were employed.

**Moments:** An average RMSE of 0.11 Nm/BW was computed for the external joint moment ($\Delta M_{\text{Joint max}}$) across all permutations. This value is within the observed range of mean error of external joint moments (sagittal plane; hip: -0.05 to 0.15 Nm/kg and knee: 0.04 Nm/kg) [164, 165] calculated using different approaches during gait, despite the fact that the magnitudes of the moments in this study were considerably higher. Here, large SDs were observed across the results for different permutations (Figure 5.4), indicating that scaling and kinematic weighting concepts could lead to a considerable variability in the levels of robustness of the calculated joint moments. Compared to RoM and JCoRs, the parameter $\Delta M_{\text{Joint max}}$ seems to be robust over many permutations, hence suggesting low sensitivity of this parameter to any specific input, as long marker and angle weighting are included; however, non-physiological kinematic simulation clearly affects the moments and any resulting MS analysis, and should be avoided at all costs.

Interestingly, the RMSEs of $\Delta RoM_{\text{Joint}}$ and $\Delta M_{\text{Joint max}}$ differ considerably with the inclusion or exclusion of weighting factors for the pre-calculated joint angles in the MS simulation (Table 5.1), even although the applied weighting factors were extremely small (0.02). Since SDs of the RMSEs of $\Delta RoM_{\text{Joint}}$ and $\Delta M_{\text{Joint max}}$ were in general large when pre-calculated joint angles or all skin markers were excluded from the kinematic weighting (Table 5.1), the accuracy of reproducing the RMD remains somewhat unknown. On the other hand, by including pre-calculated joint angles, low RMSEs as well as low SDs can be achieved for all three parameters (hence indicating high robustness). Since the outcomes of a MS simulation are highly dependent on the ability to accurately reproduce the RMD, and therefore reliant upon the scaling and kinematic weighting concepts, the final permutation chosen should be directly based on the specific aim of the study, e.g. if accurate determination of movements of a specific joint is required or kinematic or kinetic aspects are of high interest.

To consider all these aspects, a specific ranking was developed to evaluate the different weighting concepts that could then be used for future studies (Table 5.3). In our case, all permutations of the first 17 rankings included fCoRs in both the scaling and the kinematic weighting concepts. Therefore, the inclusion of fCoRs in the scaling procedure in the simulation, instead of simply using anthropometric based data, is appropriate. Permutations that only used skin marker data to scale the reference MS model and register the kinematic movement without the inclusion of fCoRs led to a rank above 55 and resulted in a $\text{Sum rel RMSE}$ of 2.75 (Norm RMSE: $\Delta d^\text{Joint} = 1.70$; $\Delta RoM^\text{Joint} = 0.09$; $\Delta M^\text{Joint max} = 0.96$) or higher. Due to efficiency reasons or if pre-calculated joint
angles are not available, rank 7 with a Sum rel RMSE of 1.37 should be used such that low errors can be achieved. Despite being ranked only 5, the solution including fCoRs, with automatic weighting of the skin markers based on STA and joint angles clearly also has positive attributes: firstly, the Sum rel RMSE is almost as low as the rankings above it. Secondly, by including all weighting possibilities, large errors can be avoided and high levels of robustness can be achieved. Most importantly, however, the automated skin marker weighting process is the most objective approach for weighting skin markers in a batch simulation process.

Two main limitations of this study should be mentioned: The chosen model was developed for gait analyses and might be inaccurate when analysing motions with high degrees of knee flexion (such as the split squats simulated in this study), which has also been noted by the authors of this particular OpenSim model [166]. However, as far as we are aware, there is no existing and validated model that is able to satisfy all these conditions. The second limitation was the lack of true gold standard evaluation data such as dual-plane fluoroscopic data combined with instrumented joint implants. Furthermore, the evaluation parameters $\Delta RoM_{Joint,p}$ and $\Delta d_{Joint}$ are directly linked to the simulation input data of the pre-calculated joint angles and the 3D location of the fCoRs, and are therefore inherently linked to the input RMD.

This study provides a guideline for scaling and weighting concepts of input data for the inverse kinematics procedure in MS simulations. Furthermore, the typical errors involved in using scaling and kinematic weighting concepts are presented. The results indicate that the scaling and weighting of reference movement data that includes fCoRs, pre-calculated joint angles, and skin markers, provides a sound basis for ensuring robust and high quality scaling of the reference MS models and consequent registration of kinematic data for the individualisation of MS models.
Declarations

Competing interests

The authors declare that they have no competing interests.

Authors’ contributions

All authors contributed to conception and design, paper preparation, read, and approved the final paper.
Chapter 6

Evaluation of the accuracy of musculoskeletal simulation during squats by means of instrumented knee prostheses

adapted from:

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In review
Chapter 6: Accuracy of Musculoskeletal Stimulation during Squats

6.1 Abstract

Standard musculoskeletal simulation software tools now offer widespread access to internal loading conditions for use in improving rehabilitation concepts or training programmes. However, despite broad reliance on their outcome, the accuracy of such loading estimations, specifically in deep knee flexion, remains generally unknown. Therefore, the aim of this study was to evaluate the error of tibio-femoral joint contact force calculations using musculoskeletal simulation compared to in vivo measured joint contact forces in a group of instrumented total knee endoprostheses patients during squat exercises.

Using scaled reference musculoskeletal models (OpenSim), tibio-femoral joint contact forces were calculated in 6 subjects for 5 repetitions of squat cycles. Tibio-femoral joint contact forces of 0.8-3.2 times bodyweight (BW) were measured. While the musculoskeletal simulations underestimated the measured knee joint contact forces at low flexion angles, an average error of less than 20% was achieved at 25°-60° knee flexion during both eccentric and concentric movement. With an average error that behaved almost linearly with knee flexion angle, an overestimation of approximately 60% was observed at deep flexion (80°). In one individual, an absolute maximum error of ca. 1.9BW was determined. Our data indicate that loading estimations from musculoskeletal loading simulations at high and low knee joint flexion angles should be considered carefully. An improved understanding of the aetiology of these potentially large errors is clearly required before targeted improvements to standardised models can occur.

Keywords

joint contact force, deep knee flexion, musculoskeletal modelling, in vivo validation, strength training
6.2 Introduction

Accurate knowledge of the internal loading conditions in the human musculoskeletal (MS) system, including muscle and joint contact forces (JCFs), can provide a strong evidence based foundation for improving rehabilitation concepts and customising training programmes as well as for optimising implant designs. Although direct, non-invasive access to muscle and JCFs is, in most cases, not possible, internal loading conditions have become widely available using MS simulation [129]. The accuracy of such simulations, especially if large ranges of motion (RoMs) are considered, is known to be sensitive to a wide variety of parameters, but often differs in a complex manner from the real *in vivo* situation [54, 92].

During strength exercises, extreme postures with large joint angles such as high knee flexion during squats is of interest due to fact that the greatest joint loading conditions are thought to occur in these postures [84, 167]. While modelling approaches to estimate these internal muscle and joint contact forces have become more widely available in a variety of software modelling packages (e.g. Anybody, OpenSim, LifeModeler™, MSIM, Biomechanics of Bodies, etc.), validation of the output of MS simulations, especially at higher angles of joint flexion, remains difficult due to limited *in vivo* data [91]. One risk is the growing reliance on knowledge gained from standard reference models for application to a wide range of subjects and activities, including rehabilitation and training exercises with high knee flexion angles, without knowing the associated validity of data and the corresponding levels of error. In selected cases, *in vivo* force measurements using instrumented total knee endoprostheses is possible [168], allowing access to the contact forces that occur across the tibio-femoral joint during a variety of activities. Such *in vivo* data may serve as a reference for evaluating musculoskeletal load analyses. Therefore, the aim of this study was to evaluate the accuracy of JCFs determined using open source MS simulation tools compared to tibio-femoral JCFs measured in 6 subjects each with an instrumented total knee arthroplasty (TKA) during squat exercises.

6.3 Methods

Six subjects (5m, 1f, aged 68 ± 5 years, mass 88 ± 12kg, height 173 ± 4cm) each with an instrumented TKA were measured while each performing five valid repetitions of a squat exercise without additional weight. Each subject possessed an INNEX knee implant (Zimmer, Switzerland; type FIXUC), in which the tibial component was instrumented with a 9-channel telemetry
transmitter (90-100 Hz), which allowed six-component load measurements of the 3 contact forces and 3 joint moments acting on the tibial component to be recorded [168]. To analyse the motion of the body, 55 skin markers were attached mainly to the lower extremities [86], and an opto-electronic system (Vicon, Oxford Metrics Group, UK) with 22 cameras (MX40 and MX160) captured the kinematics at a sampling frequency of 100 Hz. The ground reaction forces (GRFs) were measured using two force plates (type 9281B and 9287B Kistler, Switzerland), one under each foot, at a frequency of 2 kHz. All measurement systems recorded simultaneously and were temporally synchronised.

Each subject performed basic motion tasks according to List and co-workers (2013) [86] in order to functionally determine the centres of rotation (fCoRs) and axis of rotation of the hip, knee and ankle joints. The measured squat activity then consisted of each subject standing with stationary feet, approximately shoulder width apart, and hands stretched forwards. Knee joint flexion was then performed as far as possible before returning to the standing position. The kinematic and kinetic data were reconstructed in Vicon Nexus (v1.8.5, Oxford Metrics Group, UK) and further processed using in-house software written using Matlab (R2014a, Mathworks, Natick, MA, USA) to extract skin marker and joint centre locations for each time frame, as well as joint angles and GRFs. The averaged velocity of the shoulder mounted skin markers was used to define the duration of the squat cycle ($v_{shoulder} > 20 \text{mm/s}$).

The kinematic and kinetic data of the squat exercises served as input to the musculoskeletal models. To estimate tibio-femoral JCFs, MS simulation (OpenSim SimTK 3.3, Stanford, USA) of each subject and cycle was performed using the "Gait2392_simbody" standard model [50, 148, 149, 169–171] including 14 body segments, 23 degrees of freedom and 92 muscles [172]. Each model’s segment lengths were scaled based on the fCoRs of the ankle, knee and hip. Marker weightings in the inverse kinematics process were chosen as follows: hip, knee and ankle with 50; skin markers using an automated weighting procedure based on STA [146, 153, 173] with a total weight of 10 for each segment.

To calculate muscle moment arms, especially for multi-joint muscles, the OpenSim generalized force method was used, which takes wrapping and via points of muscles into account [174]. A static optimization criteria that minimized the sum of the squared muscle activation [175–177] at each time frame was used to calculate the muscle activities and forces, under the assumption that each muscle generated a length and velocity dependent force output (Hill-type muscles) [150, 172]. The low pass filter was set to 6 Hz and the maximum number of iterations was fixed.
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at 100. Total JCFs (also known as joint reaction forces within OpenSim) for each subject were calculated at the knee for each complete cycle ($JCF_{MS}$) as the sum of the calculated muscle forces, ground reaction forces, segment masses and inertial forces.

To evaluate the accuracy of the quasi-static optimization approach for determining internal forces at the knee (and indirectly the quality of the muscle force estimations), the computed JCFs of the MS simulation ($JCF_{MS}$) were compared against the measured JCFs from the instrumented TKAs ($JCF_{TKA}$). The $Error_{JCF}$ was calculated for each repetition in each subject as follows:

$$Error_{JCF}[\text{in \%}] = \left( \frac{JCF_{MS} - JCF_{TKA}}{JCF_{TKA}} \right) \times 100$$ (6.1)

Additionally, the results obtained for all 6 subjects were arithmetically averaged inter-individually and presented as a function of the knee flexion angle [$^\circ$]. Furthermore, the maximal and minimal measured ($JCF_{TKA,max}$, $JCF_{TKA,min}$) and calculated ($JCF_{MS,max}$, $JCF_{MS,min}$) forces within each cycle were extracted and used to calculate the average peak error ($Error_{JCF_{max}}$) between the simulated and measured JCFs.

6.4 Results

Joint contact forces on the tibial plateau ($JCF_{TKA}$) of between 0.8 and 3.2 times bodyweight (BW) were measured in vivo across the six patients during the squat activity (Table 6.1). The simulated forces for the same activities were calculated to be between 0.4–5.1 BW, both under- and overestimating the measured $JCF_{TKA}$ at different flexion angles. The averaged peak error ($Error_{JCF_{max}}$) over all subjects and squat cycles predicted by the MS models was 58.5\%, while the inter-subject standard deviations of this error was around 9\%. An almost linear dependence of the Error JCF was observed with knee flexion, resulting in single cycle absolute errors of up to 100\% at deep knee flexion angles (Figure 6.1). The range of knee flexion where the errors in the JCF remained between ±20\% was 26.5° - 60° during eccentric and 27.5° - 60.5° during concentric movement.
Table 6.1: Extreme (max or min) joint contact forces (JCF) of the knee measured in subjects using instrumented total knee arthroplasty (TKA) or calculated by means of musculoskeletal simulation (MS) as well as the Averaged Peak Error [%].

<table>
<thead>
<tr>
<th>Subject</th>
<th>Instrumented Implant</th>
<th>MS Simulation</th>
<th>Averaged Peak Error</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$JCF_{TKA,min}$ [BW]</td>
<td>$JCF_{MS,min}$ [BW]</td>
<td>Error$JCF_{min}$ [%]</td>
</tr>
<tr>
<td></td>
<td>$JCF_{TKA,max}$ [BW]</td>
<td>$JCF_{MS,max}$ [BW]</td>
<td>Error$JCF_{max}$ [%]</td>
</tr>
<tr>
<td>1</td>
<td>1.16</td>
<td>0.53</td>
<td>-54.1</td>
</tr>
<tr>
<td>2</td>
<td>0.86</td>
<td>0.44</td>
<td>-48.9</td>
</tr>
<tr>
<td>3</td>
<td>0.75</td>
<td>0.59</td>
<td>-21.0</td>
</tr>
<tr>
<td>4</td>
<td>0.94</td>
<td>0.87</td>
<td>-7.5</td>
</tr>
<tr>
<td>5</td>
<td>0.88</td>
<td>0.47</td>
<td>-47.3</td>
</tr>
<tr>
<td>6</td>
<td>1.34</td>
<td>0.59</td>
<td>-55.8</td>
</tr>
<tr>
<td>Av</td>
<td>0.99</td>
<td>0.58</td>
<td>-39.1</td>
</tr>
<tr>
<td>Std</td>
<td>0.20</td>
<td>0.14</td>
<td>18.2</td>
</tr>
</tbody>
</table>

Figure 6.1: Errors in the joint contact force ($ErrorJCF$, shown in %) of the simulated forces are shown compared to the telemetrically measured knee joint contact force across the full range of knee flexion achieved during the squat activity. The thin lines represent the average error in each individual subject. The thin dotted lines represent regions where not all 5 trials achieved this flexion angle. The black thick line depicts the average of all 6 subjects (averaged JCF error), where the dotted region shows the results for which not all subjects achieved this flexion angle. The grey areas show the average range of joint flexion (dashed lines; $26.5^\circ$ - $60^\circ$ eccentric; $27.5^\circ$ - $60.5^\circ$ concentric) that achieved an averaged JCF error of within ±20%.
Chapter 6: Accuracy of Musculoskeletal Stimulation during Squats

6.5 Discussion

The usage of standard reference models for the determination of internal loading conditions has become commonplace, but the accuracy of such models, particularly during more challenging activities that include higher RoMs has, until now, remained limited. In this study, to our knowledge for the first time, a reference open source MS model has been bench-marked against in vivo measured forces for six subjects throughout repetitions of the squat activity. Our data suggest a clear relationship between the error in force calculation and the knee flexion angle. At higher joint flexion angles – for example during strength exercises such as squats – prediction of internal loading conditions using MS simulation is associated with higher errors and seems to be challenging. However, with an average error range of ±20% (individual errors of up to +53% or -45%), the results of this study do indicate that the investigated MS model is indeed able to estimate knee JCFs during squats, and possibly also similar activities of daily living, where the knee joint flexion angle remains within approximately 25-60°. However, it is important to note that a large inter-subject variation was observed (Table 6.1, Figure 6.1), signifying that results of a single individual should be interpreted with care.

In strength training, the highest loading conditions are associated with positive adaptation but also possible injury, and often occur in positions of deep knee flexion [84]. However, the loading conditions in these postures are now known to be estimated with least accuracy. While a comparison of loading conditions across exercises each with similar flexion angles should still be possible, our results indicate that comparisons at different flexion angles should be interpreted with caution. The flexion dependent error also possibly explains the higher average peak errors (Error\(JCF_{\text{max}}\)) found in our study (58.5%) compared to a previous study analysing daily activities such as gait (11%) or stair climbing (26%) [91], where their activities were performed at lower knee flexion angles. However, since a low peak tibio-femoral contact force error of 14±10% was observed in their study during squatting activities at 90° flexion, it seems that the error is likely to be governed by more than just differences in joint flexion angles alone.

The aetiology of the high errors in internal loading conditions at low and high flexion angles remains unclear. While it is entirely possible that a regression relationship (\(Error\(JCF_{\%}\) = 1.3-x+57.5; where x represents the knee flexion angle; \(R^2=0.99\)), based on the data from our study could be used to correct predictions from the standard OpenSim reference model (Gait2392_simbody), it should be recognised that improvements to the underlying model should occur if possible rather than adjusting the output forces to reduce errors. The clear question...
posed by the results of our study is: why does the investigated standard MS model have less
capacity to accurately predict forces at extreme ranges of knee flexion? Here, it is entirely possi-
ble that different marker and joint weightings as input to reconstructing the skeletal kinematics
could lead to large differences in the simulated JCFs. As a result, follow-up studies should con-
sider a variety of kinematic parameters (e.g. weightings to the markers, functional joint centres,
and joint angles etc.), including improved anatomical individualisation, and the role they play
on modifying the load predictions [146, 173]. However, since the largest errors were observed
at the extreme (both high and low) RoMs, it is probably that an improvement to modelling
parameters such as the dynamic wrapping of muscles (particularly at the knee and hip [178])
and the associated dynamic change of lever arms, the inclusion of passive soft- and connective-
tissue forces (e.g. ligaments, joint capsule etc.), muscle co-contraction and goal-oriented muscle
optimisation [179] could all play key roles for achieving more accurate predictions at high and
low flexion angles.

Despite the fact that the validation of MS models using an instrumented implant is currently
the gold standard, there are indeed a number of limitations to this study. It is clear that
our results are based only on a small population of elderly subjects, using implants where the
cruciate ligaments have been sacrificed [167, 180]. In this respect, the lever arm of the patella
tendon in particular, which is known to play a critical role for the prediction of tibio-femoral
JCFs [181], might well differ from the standard MS models used for their analysis in this study.
Additionally, this elderly cohort was not experienced in strength training, lifting or even in the
squatting activity analysed, possibly explaining the large inter-subject variability observed in the
data.

This study has examined the measured and simulated internal joint contact forces in 6 subjects
throughout complete cycles of squatting. A knee joint flexion dependent error was observed
with region of high accuracy between approximately 25-60°, suggesting that widely available
and standard MS models are well equipped to provide loading predictions during activities that
involve mid-range flexion angles, but indicate that loading estimations at low and high knee joint
flexion angles should be interpreted carefully. An improved understanding of the aetiology of
these potentially large errors is clearly required before targeted improvements to standardised
models can occur.
Declarations

Acknowledgements

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Competing interests

The authors declare that they have no competing interests.
Chapter 7

Towards Evidence Based Strength Training: A Comparison of Muscle Forces during Deadlifts, Goodmornings and Split Squats

adapted from:

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In review
7.1 Abstract

To ensure an efficient and targeted adaptation with low injury risk during strength exercises, knowledge of the subject specific internal loading conditions is essential. The goal of this study was to calculate the lower limb muscles forces during the strength exercises deadlifts, goodmornings and splits squats by means of musculoskeletal simulation. 24 participants were assessed performing either 10 different variations of split squats, or deadlift and goodmorning exercises. Using individualised musculoskeletal models, forces of the Quadriceps (four parts), Hamstrings (4 parts) and Gluteus maximus (3 parts) were computed. Deadlifts resulted highest loading for the Quadriceps, especially for the vasti, but not for the rectus femoris, which exhibited its greatest loading during split squats in the rear limb. Hamstrings were loaded isometrically during goodmornings but dynamically during deadlifts. For the Gluts, the highest loading was observed during split squats, while deadlifts produced increasingly large loading over large ranges of motion. For all examined muscles, deadlifts produced considerable loading over large ranges of motion, while split squats seem to be highly dependent upon exercise variation. This study provides key information to support evidence-based strength-training design for specific muscles with respect to loading and range of motion.

Keywords

Musculoskeletal Modelling; Muscle Forces; Strength Exercises; ACL
Chapter 7: Muscle Forces during Strength Exercises

7.2 Introduction

Resistance exercises are widely used in training programs and during rehabilitation to enhance performance, health and fitness through strengthening and adaptation of specific soft tissue and musculoskeletal structures. Prevention of muscle atrophy and an increase of lean body mass, together with a decrease of body fat and improvements in body mineral density, as well as increased insulin sensitivity, have all been demonstrated as positive side effects [10]. To ensure an efficient and targeted adaptation with low injury risk, as well as a safe training design, knowledge of the subject specific loading conditions, but also a clear understanding of the external and internal kinematics and kinetics of the resistance exercises themselves, are essential.

Studies have investigated external (e.g. joint moments) [85, 87] as well as internal loading conditions (e.g. patello-femoral joint contact forces or muscle activities based on EMG measurements) [59, 71] during training exercises such as deadlifts (DLs), goodmornings (GMs) or split squats (SQs), resulting in a variety of training recommendations. However, precise and objective comparisons of the actual muscle forces that act throughout these exercises and provide a more complete understanding of the loading patterns for laying the foundations for evidence-based training recommendations, are clearly missing in the literature. Unfortunately, it is not currently possible to measure muscle forces experimentally in a non-invasive manner during exercise performance in vivo. Moreover, data from alternative measurement techniques are not yet sufficient for deducing internal forces and moments for complex dynamic systems such as the lower limbs, in a straight forward manner [45]. Computational models that are able to provide an insight into the internal loading conditions in the human musculoskeletal system [129] have become available using different software packages (e.g. OpenSim SimTK, LifeModeler™, Anybody Modeling System™, Biomechanics of Bodies). Such models are now widely used in clinical and biomechanical gait analysis for studying lower limb dynamics as well as for investigating loading conditions in strength exercises [48, 49, 75]. An improved understanding of these muscle and joint contact forces, including not only the magnitude but also the direction of the forces, is essential for appropriate prescription and modification of training exercises, as well as for improving rehabilitation outcomes [48, 180].

DLs, GMs and SQs are all multi-joint resistance exercises used to enhance athletic performance or for reducing the risk of musculoskeletal injury, but also for specific rehabilitation programmes during recovery from injury through targeted improvement of the dynamic stability of lower limb joints [33]. DLs begin with the lifter in a squat position, with arms straight and directed
downwards, with an alternating handgrip on the bar. The movement consists of an extension of the knees and hips until the body reaches an upright standing position. GMs start in an upright standing position and, with the barbell on the shoulders, the hips are progressively flexed until maximum hip flexion is reached, but the knees remain straight throughout. Similarly, SQs are performed with the barbell on the shoulders, but one foot is placed in an anterior position while the subject flexes the knees as far as possible. GMs, DLs and SQs are thought to be comparable in their ability to train strength, speed and power in all sport types [117], as well as for improving joint stability in ACL rehabilitation [118], but also with respect to their potential for injury while performing the exercises [119]. However, the individual internal loading conditions, and specifically the muscle forces that occur during these different exercises remain unknown.

Therefore, with the goal to improve training and rehabilitation monitoring and design with respect to specific joint motion, the aim of this study was to examine and compare the internal loading conditions, especially the specific muscle forces that occur in the lower limbs’ large muscle groups (Hamstrings, Quadriceps and the Gluteal Maximus), during DLs, GMs and SQs.

7.3 Methods

A total of 24 subjects, all experienced in weightlifting, were analysed while performing either DLs and GMs, or SQs. For the DLs and GMs, a total of 13 subjects (4 female, 9 male; 25 ± 4 years, 74 ± 11 kg, 1.80 ± 0.07 m) were observed [85], while five female and six male subjects (25 ± 3 years, 68 ± 9 kg, 1.76 ± 0.07 m) were analysed while undertaking SQs [87]. The local ethics committee approved both studies, and participants each provided written informed consent before commencing the testing. To analyse the motion of the body, an opto-electronic system (Vicon, Oxford Metrics Group, UK), recording at 100Hz with twelve cameras (MX40) was used. Ground reaction forces under each foot were measured using two 400x600mm force plates (type 9281B Kistler, Winterthur, Switzerland) at a frequency of 2kHz. Additionally, the force plates were calibrated to accurately determine the centre of pressure [120]. The IfB Marker Set [86] consisting of 55 skin markers, applied mainly on the lower extremities of the subjects was attached by trained personnel. Subjects wore their normal training shoes throughout the exercise testing.

After an appropriate warm up, all subjects performed standardized basic motion tasks to functionally determine the centres of rotation (fCoR) and axes in the hips, knees and ankle joints according to List and co-workers [86]. An additional load of 25% of each subject’s bodyweight
(BW) was added to the barbell (SQs, DLs and GMs) as well as 50% of the subject’s BW for DLs only. The step length of the SQs, as well as the frontal tibial angle, was varied and led to 10 different variations of SQs [87]. For each exercise and exercise variation, 6 (for SQs) or 8 (for DLs and GMs) repetitions were performed by each subject.

Captured data were further processed in Matlab (R2014a, Mathworks, Natick, MA, United States) to extract skin marker and joint centre locations in space for each time frame, as well as joint angles and GRFs. Velocity of the barbell markers was used to define the start- and end-point of the strength exercise cycle \( v_{\text{barbell}} > 40 \text{mm/s} \). Using OpenSim (SimTK, Stanford, CA, United States [49]), the extracted data were used to calculate muscle forces in the lower extremities [173]. Here, an adapted standard and widely used musculoskeletal model (“Gait2392_simbody” [50]) including 14 body segments, 29 degrees of freedom (including 3 DoFs in each knee and ankle joint) and 92 muscles [172] was scaled to each subject’s individual segment length using the fCoRs of the hip, knee and ankle joints, as well as all the skin markers and pre-calculated joint angles. Using the subject specific models, the kinematics (in OpenSim termed: inverse kinematics) were calculated as recommended by Schellenberg and co-workers [173], where the fCoRs of the hips, knees and ankles were weighted with a factor of 100, 100 and 60 respectively. Furthermore, all attached skin markers were automatically weighted based on soft tissue artefact [146, 173] with a total weighting of 10 for each segment; pre-calculated joint angles were weighted with 0.02. A standardised OpenSim static optimization, using a cost function that minimised the sum of the squared muscle activation at each time frame, was performed using 6Hz low pass filtered resultant kinematic data and the measured ground reaction force.

Quadriceps, Hamstrings and Gluteus Maximus muscle groups were evaluated. Quadriceps consisted of m. rectus femoris (RectFem), m. vastus lateralis (VasLat), m. vastus medialis (VasMed), and m. vastus intermedius (VasInt); while the Hamstrings consisted of m. biceps femoris long (BifemLh) and short head (BifemSh), m. semitendinosus (Semiten) and m. semimembranosus (Semimem). The Gluteus Maximus muscle was considered to consist of three different parts, max1 (representing the lateral part), max2 (representing the intermedial part) and max3 (representing the medial part) (Figure 7.1).

Maximal muscle forces and the muscle forces relative to the joint angles during the typical SQ (a stride length equal to 70% of subject’s leg length and a maximal tibial angle of 90°) were calculated for both the front and rear limbs, and compared against DLs with 25% and 50% additional load, as well as GMs with 25% additional load. All muscle forces were normalized to
Figure 7.1: Examined muscles, where the Quadriceps muscles are shown in the frontal plane, while the Hamstrings and Gluteus Maximus muscles are shown in the dorsal view.
Chapter 7: Muscle Forces during Strength Exercises

the subject’s BW and are thus presented in the unit N/kg (i.e. force per kg of bodyweight). Using
the averaged repetitions of each exercise, the influence of the different exercises on each muscle
of the three different muscle groups was analysed using a linear mixed model with maximal
muscle forces and exercise variation as fixed effects and subjects as random effects. Furthermore,
to examine maximal muscle forces between the 10 different SQ execution forms, a linear mixed
model with step length and tibial angle as fixed effects and subjects as random effects was used. A
Bonferroni post-hoc test was conducted in all aforementioned cases if significances were detected.
Statistical tests were performed using IBM SPSS (version 22, SPSS AG, Zürich, Switzerland).

7.4 Results

All trials of all subjects could be successfully simulated. The detailed report of the statistical
tests is presented in the online supplementary material. There was no difference between the two
cohorts regarding age, weight and height. Absolute knee joint contact forces averaged over all
subjects ranged from 9-111 N/kg for DLs with 25%; 12-127 N/kg for DLs with 50%; 14-47 N/kg
for GMs; 14-120 N/kg for all types of SQs in the front limb; 17-65 N/kg for all types of SQs in
the rear limb.

7.4.1 Quadriceps Muscles

Different exercises and the different execution form influenced the Quadriceps muscle forces.
During GMs, almost no forces in the Quadriceps muscles were observed (Table 7.1, Figure 7.2).
For DLs and the front limb of SQs, forces were similar, and higher (for VasInt and VasLat)
compared to the rear limb of SQs (Table 7.2). For both exercises, VasLat showed the highest
forces followed by VasMed, VasInt and RectFem (Table 7.1). During DLs, at knee angles of 40
°
and above, highly loaded Vasti muscles were observed (Figure 7.2). The RectFem force of the
rear leg showed a rather constant muscle force (>7 N/kg) over the whole RoM (Figure 7.2) while
the vasti forces were lower (Table 7.1).

For the different SQs, vasti muscle forces of the front limb decreased with increasing tibial
angle (Tables 7.1 and 7.3), while they increased with increasing tibial angle in the rear limb.
Additionally, the VasMed force of the front limb increased with step lengths above 55% of leg
length. A large step length with a small tibial angle (60°/75°) resulted in a higher RectFem
muscle force (Tables 7.1 and 7.3). Contrary to the front limb, VasLat and VasMed of the rear
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Figure 7.2: Muscle forces normalised to each subject’s bodyweight as a function of knee and hip flexion angles for the exercises goodmornings (GMs), deadlifts (DLs) and split squats (SQs) using a step length of 70% of the subject’s leg length and a maximal tibial angle of 90°. The four parts of the Quadriceps muscles (m. rectus femoris in blue; m. vastus intermedius in red; m. vastus lateralis in purple; m. vastus medialis in green), together with the four parts of the Hamstrings muscles (m. biceps femoris long head in blue; m. biceps femoris short head in red; m. semitendinosus in purple; m. semimembranosus in green), are shown in rotation format in blue in relation to the subject’s leg length and a maximal tibial angle of 90°. The three parts of the Gluteus Maximus muscle (max1 in blue; max2 in red; max3 in purple) are also depicted.
Table 7.1: Maximal normalized muscle forces and standard deviation of 11 muscles for 23 different strength exercise variations. GM: goodmorning; DL: deadlifts; LL: % of subjects’ leg length; TA: tibia angle relative to the ground; Added weight: Additional weight on the barbell as % of subject’s bodyweight

<table>
<thead>
<tr>
<th></th>
<th>RectFem</th>
<th>VasInt</th>
<th>VasLat</th>
<th>VasMed</th>
<th>BiFemLh</th>
<th>BiFemSh</th>
<th>Semimem</th>
<th>Semiten</th>
<th>Gluteus Maximus</th>
</tr>
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<tbody>
<tr>
<td>GM 25</td>
<td>1 ± 2</td>
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<td>6 ± 5</td>
<td>28 ± 5</td>
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<td>6 ± 3</td>
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<td>8 ± 3</td>
</tr>
<tr>
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<td>33 ± 3</td>
<td>26 ± 5</td>
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<td>28 ± 7</td>
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<td>5 ± 3</td>
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<td>26 ± 6</td>
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<td>8 ± 2</td>
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<td>SQ 25 r 70 % 75°</td>
<td>18 ± 5</td>
<td>10 ± 2</td>
<td>15 ± 5</td>
<td>15 ± 4</td>
<td>0 ± 0</td>
<td>5 ± 3</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
</tr>
<tr>
<td>SQ 25 r 70 % 90°</td>
<td>14 ± 3</td>
<td>12 ± 2</td>
<td>22 ± 5</td>
<td>19 ± 5</td>
<td>0 ± 0</td>
<td>5 ± 2</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
</tr>
<tr>
<td>SQ 25 r 85 % 60°</td>
<td>27 ± 4</td>
<td>5 ± 3</td>
<td>5 ± 5</td>
<td>7 ± 3</td>
<td>0 ± 0</td>
<td>7 ± 5</td>
<td>0 ± 1</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
</tr>
<tr>
<td>SQ 25 r 85 % 75°</td>
<td>26 ± 4</td>
<td>7 ± 3</td>
<td>8 ± 6</td>
<td>9 ± 5</td>
<td>0 ± 0</td>
<td>7 ± 4</td>
<td>0 ± 1</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
</tr>
<tr>
<td>SQ 25 r 85 % 90°</td>
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<td>11 ± 2</td>
<td>17 ± 5</td>
<td>16 ± 5</td>
<td>0 ± 0</td>
<td>6 ± 3</td>
<td>0 ± 0</td>
<td>0 ± 0</td>
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</tr>
<tr>
<td>SQ 25 r 85 % 105°</td>
<td>18 ± 6</td>
<td>12 ± 2</td>
<td>20 ± 6</td>
<td>17 ± 4</td>
<td>0 ± 0</td>
<td>6 ± 3</td>
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limb increased with an increasing tibial angle and decreased with a larger step.

### 7.4.2 Hamstrings Muscles

During all the examined exercises, the force of Semiten remained low (\(< 5 \text{ N/kg}\)) compared to other muscles, with the highest muscle force observed when performing DLs with 50\% additional weight (Table 7.1). The RoMs and the forces of the other Hamstring muscles were highly influenced by the chosen exercise. During GMs the RoM of the Hamstrings remained rather low. Furthermore, BifemSh, BifemLh and Semimem muscles were all loaded but only BifemSh produced a force that was significantly higher than during DLs and in the front limb of the SQs. In contrast, BifemLh and Semimem muscle forces were lower (Tables 7.1 and 7.2), but increased with knee flexion angles of \(> 50^\circ\) during DLs and in the front limb of SQs. The muscle forces produced by these two muscles were rather constant at knee angles between 15\(^\circ\) and 50\(^\circ\) (Figure 7.1). Forces in the Hamstring muscle group were almost negligible in the rear limb of SQs, except for BifemSh at a low knee flexion angles of around 20\(^\circ\). Here, the forces were comparable to the front limb and to DLs.

A wider stance during SQs increased the force produced by BifemSh in the rear limb (Table 7.3). In the front limb, BifemLh (over all executions: \(15.1 \pm 6.2 \text{ N/kg}\)) and Semimem (average: \(23.7 \pm 9.5 \text{ N/kg}\)) forces were considerable, since these forces were some 3 to 5 times higher than the forces produced by BifemSh. Both forces increased with an increasing step length and decreasing tibial angle (Table 7.3).

### 7.4.3 Gluteus Maximus Muscle

DLs and SQs led to significantly more force in the intermedial part compared to GMs. Furthermore, the intermedial part (\(\text{max2}\)) exhibited higher forces in all exercises (Figure 7.2 and Table 7.1) compared to the medial and lateral parts. Additional weight on the barbell during DLs only affected the medial part (\(\text{max3}\)) and therefore led to the highest muscle forces compared to all other exercises.

Similar to the Hamstrings, the Gluteal muscles were almost inactive in the rear limb of SQs, not even changing due to different execution variations. Although the value of the intermedial part (\(\text{max2}\)) increased slightly with a wider stance, the Gluteal muscle forces were relatively constant over many execution forms of the SQs (Table 7.1 and 7.3). Here, a substantial muscle
force increase with increasing hip flexion angle during SQs was observed (Figure 7.2).

7.5 Discussion

In this study, three different strength exercises, deadlifts, goodmornings and split squats, with a total of 23 different variations were analysed using musculoskeletal simulation software with the aim to compute internal loading conditions and specifically forces in the muscles. These insights are now able to form the basis for providing evidence-based recommendations for efficient training and rehabilitation.

7.5.1 Quadriceps Muscle

While the largest loading conditions in the Quadriceps were found while performing DLs, slightly lower forces were observed while executing SQs. It is therefore reasonable that DLs should be favoured if training of the Quadriceps is required, especially if the aim of the training is to strengthen vasti muscles with a desired RoM of approximately 70° to 100° in the knee. Performing SQs should be preferred if activation of vasti muscles or a smaller knee flexion angle (approximately 50° to 80°) is necessary. Since RectFem was only partially recruited in both exercises, but was constantly loaded in the rear leg of the SQs over the whole RoM (20° to 100°), the SQ exercise is recommended to strengthen this muscle, preferably using a large step length and a small tibial angle. Moreover, since loading levels of the different vasti muscles differed in both the front and the rear legs, as well as in different SQ setups (step length and tibial angle relative to the ground), an appropriate SQ setup should be considered for preventing injury and training muscular strength asymmetries within the vasti muscles.

7.5.2 Hamstrings Muscle

Since all exercises examined activated the Hamstrings but during GMs the Quadriceps remained inactive, it seems that this exercise has the ability to shift the H:Q ratio towards the Hamstrings, possibly supporting the prevention of, or recovery from, ACL ruptures [85]. However, since low RoMs in the knee were observed during GMs compared to SQs and DLs, the Hamstrings, especially the one-joint muscles such as BifemShs, will be trained only at these specific joint angles. The fact that no examined exercise seemed to be able to strengthen Semiten (as indicated
by its relatively low levels of force) indicates the requirement of other exercises to train this muscle. However, this phenomenon could result from the low maximal isometric force of the Semiten muscle compared to other Hamstrings muscles.

7.5.3 Gluteus Maximus Muscle

Schellenberg and co-workers [85] reported higher external flexion moments in the hip during GMs (1.63 ± 0.14 Nm/kg) than during DLs (1.4 ± 0.13 Nm/kg) as well as joint moments in the front limb that were comparable to SQs (1.71 ± 0.32 Nm/kg) using the same additional weight on the barbell [87]. Different to the external loading conditions of the hip in the sagittal plane, our data indicate that internal loading of the Gluteal muscles was significantly lower during GMs compared to DLs and SQs (Figure 7.2). Here, the usage of additional/other muscles seems to be a common recruitment strategy to compensate for the larger external moment [182]. This indicates that estimation of the individual muscle loading based only on the external moment might well lead to result predictions that are incorrect since joint moments alone cannot take the exercise dependent individual muscle recruitment into account. To strengthen the Gluteal muscles between a hip flexion angle of approximately 40° and 90° [87], SQs appear to require higher muscular forces and might therefore be a more efficient exercise than DLs or GMs, while DLs should be chosen if larger RoMs in the hip are required.

7.5.4 Comparison to EMG findings

Training recommendations have recently been provided based on the observed EMG activity [183, 184]. For isometric EMG measurements, the quadriceps muscle activation patterns differed across knee flexion angles and between the different muscle parts, especially VastInt, which was shown to be most sensitive to changes in the muscle length [185]. Through analysing and comparing squats, lunges, step-ups, deadlifts, and leg extensions, Ebben and co-workers [184] reported the highest activity of the biceps femoris during deadlifts. This result is only partially in agreement with our findings. Our data suggest that the short head of the biceps femoris exhibits the highest levels of activity during GMs, while the muscle force for the long head during DLs was similar to the front limb during SQs. Since the ability of EMG is known to be limited with respect to assessing the magnitude of muscle forces [185, 186], our findings clearly indicate that rating dynamic strength training exercises based purely on surface EMG measurements may be limited.
7.5.5 General

This study provides specific loading conditions during different strength exercises and therefore allows the derivation of explicit evidence-based training recommendations. The kinetic and kinematic data reported in our study were gathered during two different measurement sessions. However, no differences between the groups were observed, and all models were individualized regarding their anthropometrical data based on the basic motion tasks.

The use of musculoskeletal simulation for the determination of internal loading conditions as well as muscle activity and forces suffers from a wide variety of assumptions and simplifications. Here, the subject specific anatomy, physiology, using similar modelling techniques, Schellenberg and co-workers [187] recently showed critical errors in the use of reference musculoskeletal analyses for determining internal loading conditions during squats. An almost linear dependency of the error with knee angle was observed between measured and estimated joint contact forces, leading to an averaged maximal peak error of approximately 60% at the deepest knee flexion angles. These errors demonstrate the sensitivity of such models for predicting internal loading conditions, suggesting that the results of the current study need to be interpreted with caution. Importantly, however, the authors noted that a comparison of loading conditions across exercises with similar flexion angles should still be possible, but that comparisons across widely ranging joint flexion angles should be interpreted with caution [187]. As a result, the comparative nature of the current study should still allow a reasonable and informative comparison between the different exercises, even though the absolute magnitude of the forces reported is unlikely to be accurate. However, further investigation and validation of the force magnitudes predicted in this study is clearly indicated.

The estimated maximum stress in the muscle tissue of VasLat (muscle force per cross section area: 9.5 N/cm²) was about 61kPa during the deadlift exercise. This is higher than the commonly acknowledged ultimate stress level of muscle tissue of 50kPa [188], indicating that the estimated muscle forces are rather higher than those that actually occur physiologically. Similarly, the reported averaged joint contact forces ranged between 9 and 127 N/kg, and exceeded the in vivo measured values during deep knee bends (25 N/kg [180]) or squats (26 N/kg [187]). Compared to their cohort, our subjects lifted additional weight attached to the barbell, which is also likely to have led to additional synergistic muscle activity (mostly with a shorter lever arm) and therefore to higher joint contact forces. However, as a comparative study, the differences between the training exercises reported here still contribute new and evidence-based knowledge for informing
training and rehabilitation programmes.

One important parameter for the prevention of, or rehabilitation from, ACL rupture seems to be the H:Q ratio [23]. Under the assumption that the appropriate goal is to shift the ratio towards H, the data presented in this study suggest that GMs are a suitable exercise. However, DLs and SQs have been shown to shift the ratio rather towards Q and should therefore, in an ACL related environment, rather be avoided. If the aforementioned flexion dependent error would be taken into account, the H:Q ratio becomes even better during GMs, since this particular musculoskeletal model appears to underestimate the internal loading conditions at low knee flexion angles.

7.5.6 Practical Implications

- Muscle loading as well as ranges of motion are highly dependent on the chosen exercise and execution type.

- Deadlifts result in large ranges of motion and high loading in the thigh and pelvis muscles.

- By changing step length and angle of the frontal tibia during split squats, specific parts of the thigh and pelvis muscles can be loaded.

7.6 Conclusions

Since specific muscle forces that act during strength exercises were, until now, almost unknown, the results of this study therefore provide coaches and physiotherapists the ability to choose a minimal risk and performance targeted strength-training design for athletes or patients.
7.7 Supplementary material

Table 7.2: Significances (p < 0.05) between muscle forces of goodmornings with 25% (GM25), deadlifts with 25% (DL25) and 50% (DL50) as well as the front and rear limb of the split squats with a step length equal to 70% of subject’s leg length and a tibia angle of 90° (SQf70%90°; SQr70%90°) with 25% of the subject’s bodyweight as additional load on the barbell. Quadriceps (quad), including vastus lateralis (vl), intermedius (vi), medialis (vm) and rectus femoris (rf); Hamstrings (ham) including biceps femoris short (bs) and long (bl) head, semimembranosus (sm) and semitendinosus (st); Gluteal muscles (glut), including three different parts, max1 (g1, represents lateral part), max2 (g2, represents intermedial part) and max3 (g3, represents medial part) were examined.

<table>
<thead>
<tr>
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<th>DL 25</th>
<th>DL 50</th>
<th>SQf70%90°</th>
<th>SQr70%90°</th>
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<tr>
<td>rf, vi, vl, vm</td>
<td>rf, vi, vl, vm</td>
<td>vi, vl, vm</td>
<td>rf, vi, vl, vm</td>
<td>quad GM25</td>
</tr>
<tr>
<td>bs, sm, st</td>
<td>bl, bs, sm, st</td>
<td>bl, bs, sm</td>
<td>bl, bs, sm</td>
<td>ham GM25</td>
</tr>
<tr>
<td>g1, g2, g3</td>
<td>g1, g2, g3</td>
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<td>g2</td>
<td>glut GM25</td>
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<td>vi, vl</td>
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<td>bl, sm, st</td>
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<td></td>
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<td>g1, g3</td>
<td>g1, g2, g3</td>
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<td></td>
<td></td>
<td></td>
<td>g1, g2, g3</td>
<td>glut SQr70%90°</td>
</tr>
</tbody>
</table>
Table 7.3: Significances (p < 0.05) between muscle forces of different split squats with different step lengths (55%, 70% and 85% of subject’s leg length) and tibial angles (60°, 75° and 90° tibial angle relative to ground) for the front (f) and rear (r) limb. Quadriceps (quad) including vastus lateralis (vl), intermedius (vi), medialis (vm) and rectus femoris (rf), Hamstrings (ham) including biceps femoris short (bs) and long (bl) head, semimembranosus (sm) and semitendinosus (st), as well as the Gluteal muscles (glut) including three different parts, max1 (g1, represents lateral part), max2 (g2, represents intermedial part) and max3 (g3, represents medial part) were examined. Interactions were observed between g3 of the front limb and vi, sm, st, g1, g2 and g3 of the rear limb.

<table>
<thead>
<tr>
<th>SQ f: Step Length</th>
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<th>85 %</th>
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<tbody>
<tr>
<td>vm</td>
<td>vm</td>
<td>quad 55 %</td>
</tr>
<tr>
<td>bl, sm</td>
<td>bl, sm, st</td>
<td>ham 55 %</td>
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<tr>
<td>g2</td>
<td>g2</td>
<td>glut 55 %</td>
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<td>sm, st</td>
<td>ham 70 %</td>
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<td></td>
<td>glut 70 %</td>
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<th>SQ r: Step Length</th>
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<th>85%</th>
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<td>rf, vl, vm</td>
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<td>bl, bs</td>
<td>bl, bs</td>
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<td></td>
<td>glt 55 %</td>
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<td>rf, vl, vm</td>
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<td>ham 70 %</td>
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<td></td>
<td>glt 70 %</td>
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<tr>
<th>SQ f: Tibia Angle</th>
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<th>90°</th>
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<tbody>
<tr>
<td>vi, vm</td>
<td>vi, vm</td>
<td>quad 60°</td>
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<tr>
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<td>ham 60°</td>
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<td></td>
<td>glut 60°</td>
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<td></td>
<td>quad 75°</td>
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<td>ham 75°</td>
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<td></td>
<td>glut 75°</td>
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<th>SQ r: Tibia Angle</th>
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<td>rf, vl, vm</td>
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Chapter 7: Muscle Forces during Strength Exercises

Declarations

Competing interests

The authors declare that they have no competing interests.

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Ethical declaration

All authors abide by the Ethics Committee of ETH Zurich, Switzerland ethical rules of disclosure.

Authors’ Contribution

All authors contributed to conception and design, paper preparation, editing and approval of the final manuscript.

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8.1 Discussion

Strength training provides the ability to reduce the risk of injuries such as ACL ruptures or to eliminate existing muscular imbalances, but also to enhance specific sport performance. For all the purposes of strength training, external and internal loading conditions such as joint moments but also muscle forces are relevant parameters to ensure positive adaptations on targeted soft tissues with low risk of injury. Recent research reviewed the epidemiology of injuries across different weight training designs such as CrossFit, bodybuilding but also conventional strength training itself [189]. A relative low risk of injury with an incidence rate of 2–4 injuries per 1000 hours strength training was observed, which is lower compared to other team sports such as football, rugby and cricket (approximately 15–81 injuries per 1000 hours, [190–192]). Compared to other non-contact sports, e.g. modern dancing (approximately 0.5 injuries per 1000 hours, [193]), the injury rate of strength training remains high. Injuries and inflammations caused by strength training occur, mainly due to overloaded conditions [14]. To modulate the risk of injury, extrinsic factors such as exercise technique need to be conducted [189]. To reduce the risk of musculoskeletal injuries or in a rehabilitation process Shaw and co-worker (2016) [194] recently suggested in their literature review the usage of all strength training designs such as conventional, power, hypertrophy as well as endurance strength training designs. However, identification of important parameters which lead to overuse, and should therefore be considered to reduce risk
of injury, are still missing. Therefore, since these loading conditions are mostly unknown during many strength exercises, injury prevention programs and training recommendations as well as monitoring is not evidence based but still advised based on experience of experts such as coaches, therapist or the athletes themselves. While scientific knowledge of loading conditions, especially acting muscle forces, during strength exercises is essential to provide evidence based statements for training and therapy recommendations, it offers the ability to train in a more targeted, specific and efficient way with reduced risk of injury. The presented work closes the gap in knowledge for deadlifts, goodmornings, and split squats and further discussed the results with squats and back extensions. For these exercises, external loading conditions were analysed and by using subject-specific musculoskeletal model, individual loading conditions were calculated and evaluated. In addition, different scaling and kinematic weighting concepts were assessed in order to optimise robustness and reduce errors between the reference motion data and the musculoskeletal simulation. Furthermore, a proper validation of the musculoskeletal model was performed.

The relevant existing literature on computational techniques to determine muscle forces in the lower limb during strength exercises \textit{in vivo} and their potential for uptake into rehabilitation and sport purposes was reviewed in chapter 2. The systematic literature search resulted in 21 scientific articles (including additional papers found within the references) that met the given requirements. Different numerical techniques have been found to be reasonable determining internal loading conditions. Firstly, forward dynamics analysis was conducted to examine dynamic ballistic movement exercises such as squat jumps. Secondly, quasi static inverse dynamics optimisations were used to study low speed, non-impact movements such as split squats or leg press and thirdly, alternative methods such as EMG-driven modelling were found to estimate acting muscle forces during low speed as well as impact related exercises such as sidestepping or dynamic lunge activities. Computational numerical musculoskeletal simulation provides the ability to model subject specific internal loading conditions, especially muscle forces, during exercising. Recently, different studies investigated automatism and calibration methods for musculoskeletal modelling software but also standard data sets for model validations:

- Mantoan and co-workers [195] published a Matlab based toolbox to automatically process motion data from a c3d file into pre-processed movement data, which is used in musculoskeletal modelling software (MOtoNMS).
- Pizzolato and co-workers (2015) [196] investigated in EMG calibration methods and shared
their toolbox in an open source (CEINMS).

- Wojtusch and co-workers [197] are providing a database with high quality biomechanical measurement data combined with anthropometrical parameters (HuMoD-A).

- Ku and co-workers (2015) established the foundation for researchers to share their datasets between four different research cores [198] (the mobilizer center).

This indicates the widely spread use of musculoskeletal simulation software and their benefits to compute internal variables such as muscle dynamics or neural signals but also the importance of an automated and accurate procedure as well as the validation possibilities of musculoskeletal models.

Beside the absence of knowledge of internal loading conditions during many strength exercises, validation of the models remains lacking in literature, especially during movements including large RoM in hip and knee joint. This is related to the fact that experimental solutions measuring internal forces, stress and strains in vivo are limited. Therefore, both, investigation on acting muscle forces during exercises but also the validation of musculoskeletal models and simulations is required.

In order to provide evidence based rehabilitation and training recommendations during the strength exercises but also to make data for further musculoskeletal simulation to estimate muscle forces available, kinematic and kinetic data was measured and analysed in chapter 3 and 4. By means of a quasistatic inverse dynamic and marker cloud approach, external net joint moments with their corresponding joint angles were computed for deadlifts as well as goodmornings (chapter 3), for split squats (chapter 4) and back-extensions (chapter 4). Different loading and movement pattern were observed across all studies. Dependent on the required loading conditions as well as on the training goal, appropriate exercise should be chosen. As an example to strengthen hip and knee extensors, both over large RoMs, deadlift should be favoured compared to goodmornings. Since additional weight on the barbell during deadlift does not influence the knee moment, and higher loading on the hip with lower loading on the knee compared to squatting were observed, deadlifts should be performed to strengthen hip and back muscles. During split squats, three key findings were observed. Firstly, loading conditions and RoMs in knee and hip vary over execution forms. These results demonstrate the importance of choosing an appropriate split squat variation to target the training goal. Secondly, no evidence was found to support the common accepted guideline to prevent the knees from moving out beyond a vertical
line over the toes. Thirdly, the rear limb is loaded during split squats and should be taken into account in training designs. In order to strengthen hip and lower back, performing backextensions with a dynamic hip seems to be more efficient and healthier variation compared to the dynamic spine execution form. More efficient due to the fact that, higher moments with higher or similar RoMs in hip and lower back during dynamic hip executions was observed. Healthier because disk deformation, ligament and spinal loading can be reduced if back-extension exercises are performed with neutral lordosis [199], which was observed during the dynamic hip execution form.

To provide a more integral overview for coaches and therapist, comparison between different strength exercises using the same measurement and analyse technique is required. Therefore, results of examined exercises (deadlifts, goodmornings, split squats, squats and back-extensions) were combined. Figures 8.1, 8.2 and 8.3 provide external net moments of knee, hip and back with respect to the corresponding knee and hip angle as well as lumbar curvature of mentioned strength exercises. For deadlifts, 25% and 50% BW extra load on the barbell was used. Goodmornings and split squats were examined using 25% BW extra load only to avoid overloading. Split squat execution form was set to a tibia to floor angle of 90° and a step length of 75% of subject’s leg length, while front and rear limb of split squats were analysed separately. During squat exercises, unrestricted and restricted variations (Knees are not allowed to go beyond vertical plane over the toes) using 25% and 50% BW extra load were analysed. Back-extension exercises were evaluated using a dynamic spine and a dynamic hip variation, both performed as single and double legged execution variations.

While specific sports athletes, prevention, and rehabilitation programs need definite loading conditions as well as movement patterns in different joints, these figures provide required information to coaches or athletes. These figures enable them to choose appropriate and targeted exercises. Following key findings are pointed out explicitly: To reduce risk of ACL rupture injury, shifting of the hamstrings to quadriceps ratio towards hamstrings seems to play a key factor [23, 37–39]. Here, goodmornings or one legged Back Extension with a dynamic hip execution form are reasonable exercises and should preferentially be chosen before other examined exercises, since both knee flexors and hip extensors are loaded. To shift the ratio towards quadriceps, split squats with a focus on the rear limb should be favoured, because here knee extensors and hip flexors are loaded. Loading conditions during back-extensions in knee, hip and back are rather low compared to the other examined exercises. Here, the use of extra load on the neck could probably increase these external moments.
Figure 8.1: Averaged normalized knee moments in the sagittal plane [Nm/kg] (positive for external knee flexion moment) with corresponding knee flexion angle [°] for goodmornings (green, [85]), deadlifts (cyan, solid: 25% BW extra load, dashed: 50% BW extra load, [85]), split squats using a tibia to floor angle of 90° and a step length of 75% of subject’s leg length for the front (purple, [87]) and rear (dashed purple, [87]) limb, squats (red: unrestricted, orange: restricted, solid: 25% BW extra load, dashed: 50% BW extra load, [84]) as well as back extensions (yellow: dynamic hip, magenta: dynamic spine, solid: two legged, dashed: single legged, [200]).

*: Starting point of the eccentric phase.
Figure 8.2: Averaged normalized hip moments in the sagittal plane [Nm/kg] (positive for external hip flexion moment) with corresponding hip flexion angle [°] for good mornings (green, [85]), deadlifts (cyan, solid: 25% BW extra load, dashed: 50% BW extra load, [85]), split squats using a tibia to floor angle of 90° and a step length of 75% of subject’s leg length for the front (purple, [87]) and rear (dashed purple, [87]) limb, squats (red: unrestricted, orange: restricted, solid: 25% BW extra load, dashed: 50% BW extra load, [84]) as well as back extensions (yellow: dynamic hip, magenta: dynamic spine, solid: two legged, dashed: single legged, [200]).

*: Starting point of the eccentric phase.
Figure 8.3: Averaged normalized back moments (level L4/L5) in the sagittal plane [Nm/kg] (positive for external back flexion moment) with corresponding lumbar curvature [m$^{-1}$] for good-mornings (green, [85]), deadlifts (cyan, solid: 25\% BW extra load, dashed: 50\% BW extra load, [85]), squats (red: unrestricted, orange: restricted, solid: 25\% BW extra load, dashed: 50\% BW extra load, [84]) as well as back extensions (yellow: dynamic hip, magenta: dynamic spine, solid: two legged, dashed: single legged, [200]).

*: Starting point of the eccentric phase.
Within the last two years (2015 and 2016), many other studies investigated external loading conditions during the strength exercise squats [201–205]. Here, different techniques as well as training equipment, such as shoes, influence loading conditions and movement pattern. Examined results seem to be similar to our findings (Novice loaded squat movement with 25% BW extra loading: knee moment ranges from 0.83 – 1 Nm/kg, hip moment ranges from 1.28 - 1.43 Nm/kg [203]), and change within different cohorts [203], training equipment [201, 203] and movement patterns [201, 202, 205]. The different results within and between studies indicate the importance of a standardised procedure to compare results between studies. Furthermore, outcomes are highly related to the specific measurement setup.

Loading dependent biological adaptations are dependent on internal loading conditions such as acting muscle forces, even though analysed external moments do not necessarily reflect these internal conditions [73, 206–208]. Hence, further investigations regarding muscle loadings are required, e.g. by using computational musculoskeletal simulations.

Due to lacking validation of the musculoskeletal modelling outcomes, chapter 5 investigated in the robustness of standard musculoskeletal simulations using 90 different permutations by altering weighted scaling and kinematic input procedures during 10 different variations of split squats. Although only recently introduced, evaluation of musculoskeletal models by means of robustness is in line with suggestions on verification and validation suggestions of musculoskeletal models from Hicks and co-workers (2015) [209].

Two main key aspects are introduced in chapter 5 and are important to determine robustness of a simulation. Those are similar to novel published open access toolboxes (MOtoNMS [195], CEINMS [196] or the mobilize center [198]). Firstly, an automated procedure to write OpenSim specific file formats out of C3D (Coordinate 3D) motion data is essential. This includes ”sto” (storage), ”mot” (motion) or ”xml” (Extensible Markup Language) files. Secondly, an automated and objective procedure to scale and run the simulation is required. Here, an automated scaling based on subject specific segment lengths as well as an automated simulation technique was introduced. Three main functions can be included in the scaling and simulation process: First, the fCoRs; second, an automated skin marker weighting based on STAs (which objectively weight each marker individually with a minimized error of soft tissue movements); third, precalculated joint angles. Using the mentioned developed pipeline, split squat data set was simulated using 90 different weighted scaling and simulation permutations. Using calculated robustness, this investigation provides guidelines for registering musculoskeletal models to reference kinematic
and kinetic data. Based on the RMS errors of kinematic and kinetic parameters, a ranking for the scaling and kinematic weighting concept revealed the usage of fCoRs, automated skin markers as well precalculated joint angles in the simulation procedure for optimal registration of the musculoskeletal model to the reference motion data. This is especially essential during movement including large RoMs occurring during strength exercises such as split squats.

As stated in the literature review (chapter 2), internal loading condition validation of standardized musculoskeletal models during strength exercises remains lacking. Despite broad reliance on their outcome, the accuracy of such loading estimations remains generally unknown, especially during strength exercises where large flexion angles are considered. Therefore, chapter 6 investigated the evaluation of the accuracy of musculoskeletal simulation by means of internal \textit{in vivo} measured joint contact forces using instrumented knee prostheses of 6 subjects performing squat movements. Using scaled reference OpenSim models, calculated tibio-femoral joint contact forces were computed and compared to measured contact forces using instrumented total knee arthroplasties. An average maximal peak error of approximately $60\%$ overestimated calculated joint contact forces at deep flexion ($80^\circ$), while the simulation underestimates the measured contact forces during low knee flexion angles (upright posture). The error behaved almost linearly with knee flexion angle ($R^2 = 0.99$; Figure 6.1). Therefore, reasonable results with an error of $\pm 20\%$ or less was achieved between approximately $25^\circ$-$60^\circ$ knee flexion during both eccentric and concentric movement. Since highest loading during strength training exercises occur mostly in extreme postures and are of highest interest [84, 167], but it was observed to be the most inaccurate positions. Therefore great care should be considered by interpreting loading conditions examined using musculoskeletal simulation. Furthermore, the results indicate the need of further development of the models itself, but also an improved understanding of the "aetiology" itself. Recent investigations focused on validation of musculoskeletal models. It seems that multibody dynamics [209], musculoskeletal geometry (including modelling choices such as bodies, joints, muscle geometry) [209–211], muscle-tendon dynamics [209, 212, 213], as well as contact and other external forces [209] play a key role in musculoskeletal simulation and their outcomes. Serrancoli and co-workers (2016) [213] suggest the usage of calibrated models using \textit{in vivo} measured joint contact forces during walking compared to standard reference models. Measured joint contact forces are rarely available and the calibration method is not usable for many researchers. However, the findings regarding changes in muscle-tendon moment arms between the two models demonstrated the importance of accurate real world representations. The authors [213] especially stated the "domino-effect", which occurs by changed moment arms and
therefore changed ratios of force production between biarticular and uniarticular muscles, resulting in significantly different calculated JCFs for the same movement. Importance of other parameters is discussed controversially in literature or does not seem to have a major influence on modelling outputs. Here, influence of different numerical methods for simulation and analysis (different cost functions, such as muscle optimization techniques) on modelling outputs are inconsistent [209, 214], while general segment mass uncertainties [211] seem to have limited effects on simulations.

The reported joint contact force error in our study is rather high compared to other studies [91, 210]. Cheng and co-workers (2016) [210] reported reasonable predictions of JCF during walking using a scaled reference model with weaker maximal isometric muscle forces. Maximal isometric muscle forces are directly linked to physiological cross-sectional area of muscles. Changes in the physiological cross-sectional area and muscle tension had an influence on the rectus femoris of around 260 N [215]. In our study, computed JCFs were underestimated using lower knee flexion angles. Since weaker models required the activation of other agonist muscles – mostly with lower muscle moment arms – to compensate the same external moment, the computed JCFs in the knee are higher compared to normal strength models. Hence, weaker models lead to a shift of the accurate region shown in Figure 6.1 towards lower knee flexion angles and therefore towards flexion angles used during walking.

The high number (7) of published and here discussed studies between 2015 and 2016 shows the importance of accurate and validated musculoskeletal models and simulation techniques and can help to identify the aetiology leading to large errors reported in our study. Therefore, further research should not only address the error of calculated JCFs but also focus on accurate representation of internal anthropometric data. Focus should be taken on muscle paths and bone geometry [210, 211] including their moment arms or segment length, accurate estimation of tendon rest [212] and muscle-fibre length [212, 213], skin markers and joint weightings in the simulation process to represent reference motion data [173] to better identify the aetiology [209].

Many different published studies examine the validation of musculoskeletal models and address the need of validated and accurate musculoskeletal models, either using in-vivo [210, 211, 213, 214, 216] or in-vitro [217] measured data, or by means of a strength training related movement [218, 219] without having a gold standard comparison. However, no study was found considering both, strength exercises as well as synchronised measured in vivo tibio-femoral joint contact forces to validate musculoskeletal models. Despite all these limitations and large knee flexion
angle dependent errors found in our study, a comparison of loading conditions across different strength exercises each with similar flexion angles is still possible.

To ensure adequate tissue adaptation with low risk of injury, specific internal loading conditions such as muscle forces of each exercise should be known. In chapter 7, acting muscle forces during the strength exercises deadlifts, goodmornings and split squats were analysed with the aim to provide evidence based, safe, and efficient training programmes to coaches, therapist or athletes. Since maximal peak errors between measured and simulated joint contact force of approximately 60% were found in chapter 6, acting muscle forces were treated with caution and only compared respectively. Results indicate that biomechanical strength training relevant parameters such as maximal acting muscle forces or joint RoMs are highly dependent on the chosen exercises, especially during split squats. Deadlifts produce considerable loadings over large RoMs, while goodmornings should be favoured to prevent and rehabilitate from an ACL rupture. Recently, a few studies have been published addressing musculoskeletal simulation using a movement related to strength training [220, 221]. While Hu and co-workers (2015) [220] stated the high potential of musculoskeletal simulations to improve understanding of exercise effects in patients with specific disorders, Kolditz and co-workers [221] used musculoskeletal simulation to optimise training load with an industrial robot and an isokinematic leg extension approach. Here, the potential to develop new exercises using robots or cable cord machines to address specific muscle loading in specific RoMs is huge. The current study only taps the full potential to a small part [221].
8.1.1 Limitations

The main drawback is the accuracy of computational simulation using a standard musculoskeletal model. Accuracy was validated using instrumented total knee arthroplasties and correlates linearly with the corresponding knee angle. Calculated muscle forces need to be treated with caution and only relative comparisons between similar movement patterns are valid. Improvement of the model, especially regarding the muscle geometry path, muscle properties and moment arms is highly recommended. To improve model accuracy, subject specificity is required, especially if particular diseases or injuries should be addressed.

External moments are analysed and discussed to its entirety within this thesis and provide evidence based recommendations. However, this work has shown, that external moments, do not necessarily represent internal loading conditions such as muscle forces or ligament stress. Internal loading conditions are dependent on coordinative aspects, co-contractions of muscles to enhance stability, physical conditions of the subjects and many other factors, but difficult to measure or estimate. Since muscle forces are currently not measurable \textit{in vivo}, computational simulation needs to be conducted. Using computational simulations, aforementioned subject specific dependencies such as co-activated muscles are hard to determine using motion capture systems only. Therefore, estimation of internal loading conditions by means of musculoskeletal simulation is always subjected to risk, especially if only kinematic, kinetic and EMG data is used.
8.2 Outlook

Using musculoskeletal modelling, simulated loading conditions provide fundamental, non-invasive and accurate data. Accuracy during strength training and using a standard musculoskeletal model was shown to be limited. Different improvements are required and should be targeted:

- Improving parameters of the reference model is suggested. Fundamental improvements seem to accurately represent internal anthropometric data such as muscle paths and wrapping their moment arms over the whole RoMs but also the tendon-rest and muscle-fibre length. Models can be adapted using generic available anthropometric data sets. Subject specific modelling provides both identifying of adequate strength exercises but also preventative training designs that address subject specific injury related risk factors.

- Validate available musculoskeletal models using the data set of measured knee joint contact forces in chapter 6. Different models, such as the Arnold-model [135] or the hip musculoskeletal model [222] are freely available online. More accurate models allow a better understanding of internal loading conditions and therefore better training and rehabilitation recommendations.

- Similar to MOTOAMS [195], CEINMS [196], HuMoD-A [197] and the mobilizer center [198] (free databases containing motion capture and anthropometric data), the developed pipeline in this thesis should be an enabled open source. The pipeline includes firstly the conversion of c3d to OpenSim required files structures (mot / sto) for kinematic and kinetic data. Secondly, it provides the required setup files (xml) to scale and run a simulation. Thirdly, the procedure includes an automated scaling method based on fCoR. Fourthly, the pipeline provides an automated simulation procedure that includes different captured and analysed motion data (automated marker weighting procedure based on STAs, the inclusion of fCoR and precalculated joint angles). Sharing the pipeline improves quality and quantity of musculoskeletal simulations among different research groups.

- Measured data should be integrated in open access centers to enable a machine learning future using big data. Sharing is time intensive and not directly publishable, but benefits the research field. It allows researchers to reproduce, extend and continue but also to verify and validate other models [209].
8.3 Possible transfer to Clinic and Industry

Using the results presented in this thesis, four major transfers to industry are identified by the authors:

- Subject specific models including bone geometry and muscle paths can be created. Designing such subject specific models is time and cost intensive, but could help coaches or therapist to choose appropriate training designs. Since top athletes are nowadays fully screened anyway and money is available in many sports, clubs or athletes could have an interest in a subject specific model, especially if no additional effort by the clubs or athletes is needed. The chances to enhance sport performance with a decreased risk of injury are high. Similar, screenings of patient is mostly available in hospitals and could be used to create a subject specific model and help the patient to design rehabilitation trainings or surgeons to estimate outcomes of an intervention design required operation techniques.

- Once a subject specific model is created, evidence based requirements of an athlete or patient can be evaluated by means of musculoskeletal simulation. Such requirements are useful to design sport and rehabilitation trainings that specifically address the needs of an athlete or patient. Specific training designs could help to improve the sport performance but also to decrease the risk of injury due to better physical conditions of the athletes. To design such specific trainings, knowledge of both musculoskeletal simulation as well as the sport / rehabilitation requirements is important.

- Using the technique to systematically vary the artificial kinetics by a given kinematic, possible new or adapted exercises could be identified to specifically train a targeted muscle. Such technique includes for example a cable machine, where the subject performs a hip flexion-extension movement and the direction of the cable force acting on the ankle is varying around the shank. Using musculoskeletal simulation, thousands of movements and loading conditions could be simulated in a relative short period of time. Since the physical requirements are dependent on the sport / rehabilitation process, new or adapted exercises could be designed in each field to address these specific requirements.

- The designed pipeline to automate the musculoskeletal simulation process out of motion capture data can be certified. Using this certification, hospitals or general gait labs could be appointed with the software, which would allow routine musculoskeletal analysis in an accurate and efficient manner.
8.4 Conclusion

Strength training is used to enhance sport performance but also to prevent or rehabilitate from injuries. The focus of this work was to assess and evaluate loading conditions during different strength exercises and to recommend training as well as prevention and rehabilitation relevant suggestions. External net moments in knee, hip and lower back during different strength exercises (deadlifts, goodmornings, split squats and back-extensions) with corresponding joint angles were analysed and discussed. In this thesis, it was demonstrated that loading conditions not only vary from exercise to exercise but also within different execution forms. The results provide an evidence based fundament to choose appropriate and targeted strength exercises, but net joint moments not necessarily represent internal loading conditions such as muscle forces. Since muscle forces are currently not measurable in vivo, internal loading conditions were estimated using a musculoskeletal modelling approach. The musculoskeletal model was validated by means of instrumented total knee arthroplasty in six subjects while squatting. Maximal averaged peak error between calculated and measured knee joint contact forces of approximately 60% were found. This error indicates two key points: 1) that the reference models need to be improved to accurately estimate internal loading conditions during deep flexion angles and 2) that the estimated internal loading conditions should be interpreted with caution. To address this caution, only relative comparisons between similar strength exercises with respect to motion and external loading is suggested. Using the validated reference model and by taking the limitations into account, muscle forces acting during different strength exercises were estimated. Using estimated muscle forces acting during different strength exercises, this thesis provides and discusses evidence based training and rehabilitation recommendations with a focus on ACL injury prevention and rehabilitation.
References


References


References


References


References


References


References


References


References


Appendix A

Corresponding figures and tables of split squats and back extension exercises study (Chapter 4)
A.1 Corresponding figures and tables of split squats study (Chapter 4.1)

![Figure A.1: The measurement setup for split squats included 1) an angle meter and 2) a screen that projected a side view of the subject captured using 3) a video camera. This surveillance allowed the subject to control the angle of the front tibia. 4) Individual marks designating the step length were placed on the force plates [87].](image-url)
Figure A.2: Relation between average knee/hip angles [°] and knee/hip moments [Nm/kg] for each step length and front tibia angle [87].
Figure A.3: Overview over all conditions. Top: Knee and hip moment [Nm/kg] of the front and rear leg and the corresponding flexion respectively extension angles. Bottom: Weight distribution as vertical ground reaction force of the front foot divided by the vertical ground reaction force of both feet over a repetition [87].
Figure A.4: Relation between average tibia angle of the front leg [°] and knee moment [Nm/kg] of the front leg (top) as well as the rear leg (bottom) for each step length and front tibia angle. Positive moment represents external flexion moment in the knee [87].
Table A.1: Standardized instructions for split squat performance [87].

<table>
<thead>
<tr>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Put the barbell on the trapezius muscle, and hold it comfortably in your hand.</td>
</tr>
<tr>
<td>2 Stand upright with your feet on the hip-wide position marks. Step forward with the right leg to the specified marked step length.</td>
</tr>
<tr>
<td>3 Start with both knees in full extension.</td>
</tr>
<tr>
<td>4 Lunge as deeply as possible without touching the ground with the rear leg.</td>
</tr>
<tr>
<td>5 The specified tibia angle should be achieved at the deepest position of the split squat.</td>
</tr>
<tr>
<td>6 Keep the upper body erect during the exercise.</td>
</tr>
<tr>
<td>7 Hold tension in the core muscles during the execution of the split squat.</td>
</tr>
<tr>
<td>8 Perform the downward and upward movements of the split squat at the same normal speed.</td>
</tr>
</tbody>
</table>
Table A.2: Range of motion (ROM [°], positive for flexion) for the ankle, knee, and hip joints of the front and rear legs during split squats with different step lengths. Split squat step lengths of \( l_1 = 55\% \text{ leg length (ll)} \), \( l_2 = 70\% \text{ ll} \), and \( l_3 = 85\% \text{ ll} \) and four tibia angles (a) were evaluated. Significant differences in the ROMs for different step lengths and tibia angles are marked with a star. For the ROMs of the hip of the front and rear legs, there were significant interactions of the tibia angles and step lengths. Therefore, a Bonferroni corrected pairwise comparison could not be performed [87].

<table>
<thead>
<tr>
<th>Joint</th>
<th>Angle</th>
<th>Step Length l1</th>
<th>Step Length l2</th>
<th>Step Length l3</th>
</tr>
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<td>Ankle front leg</td>
<td>a 60°</td>
<td>41.2 ± 3.4</td>
<td>32.8 ± 3.1</td>
<td>22.8 ± 4.4</td>
</tr>
<tr>
<td></td>
<td>a 75°</td>
<td>43.4 ± 3.9</td>
<td>34.1 ± 3.1</td>
<td>22.9 ± 3.7</td>
</tr>
<tr>
<td></td>
<td>a 90°</td>
<td>45.1 ± 2.8</td>
<td>36.4 ± 2.2</td>
<td>25.1 ± 3.1</td>
</tr>
<tr>
<td></td>
<td>a 105°</td>
<td>45.1 ± 2.8</td>
<td>36.4 ± 2.2</td>
<td>25.1 ± 3.1</td>
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</tbody>
</table>

<table>
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<tr>
<th>Ankle rear leg</th>
<th>Angle</th>
<th>Step Length l1</th>
<th>Step Length l2</th>
<th>Step Length l3</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>a 60°</td>
<td>11.6 ± 6.7</td>
<td>10.9 ± 9.4</td>
<td>12.4 ± 7.9</td>
</tr>
<tr>
<td></td>
<td>a 75°</td>
<td>11.6 ± 8.2</td>
<td>12.2 ± 9.6</td>
<td>13.5 ± 9.2</td>
</tr>
<tr>
<td></td>
<td>a 90°</td>
<td>14.8 ± 6.7</td>
<td>18.8 ± 9.6</td>
<td>20.7 ± 10.3</td>
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<tr>
<td></td>
<td>a 105°</td>
<td>14.8 ± 6.7</td>
<td>18.8 ± 9.6</td>
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<td></td>
<td>a 60°</td>
<td>91.1 ± 6.9</td>
<td>82.6 ± 7.8</td>
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<td></td>
<td>a 75°</td>
<td>91.9 ± 6.6</td>
<td>81.8 ± 6.5</td>
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<td>a 90°</td>
<td>95.7 ± 8.8</td>
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<td>95.7 ± 8.8</td>
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<th>Angle</th>
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<th>Step Length l2</th>
<th>Step Length l3</th>
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<td>64.3 ± 6.8</td>
<td>69.8 ± 11.6</td>
<td>80.3 ± 11.7</td>
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<td>a 90°</td>
<td>43.6 ± 10.8</td>
<td>50.1 ± 11.9</td>
<td>62.6 ± 7.0</td>
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<th>Step Length l1</th>
<th>Step Length l2</th>
<th>Step Length l3</th>
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<tr>
<td></td>
<td>a 60°</td>
<td>42.8 ± 8.7</td>
<td>46.0 ± 10.3</td>
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<th>Angle</th>
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<th>Step Length l2</th>
<th>Step Length l3</th>
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<tr>
<td></td>
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<td>19.7 ± 6.6</td>
<td>26.3 ± 9.3</td>
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<td></td>
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<td>9.1 ± 4.7</td>
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<tr>
<td></td>
<td>a 90°</td>
<td>10.7 ± 2.7</td>
<td>7.0 ± 2.8</td>
<td>10.2 ± 4.1</td>
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149
Table A.3: P-values of the linear mixed model of the two parameters step length and tibia angle as well as their interaction. The constant term of the linear mixed model was highly significant (P-value<0.001) for all ROMs and moments [87].

<table>
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<th>tibia angle</th>
<th>interaction</th>
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<tr>
<td>ankle front leg</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>0.63</td>
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<td>ankle rear leg</td>
<td>&lt;0.001*</td>
<td>&lt;0.102</td>
<td>0.398</td>
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<td>knee front leg</td>
<td>0.221</td>
<td>&lt;0.001*</td>
<td>0.746</td>
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<td>&lt;0.001*</td>
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<td>hip front leg</td>
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<td>0.023*</td>
<td>0.020*</td>
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<tr>
<td>hip rear leg</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
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<th>tibia angle</th>
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<td>&lt;0.001*</td>
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<td>0.038*</td>
<td>&lt;0.001*</td>
<td>0.027*</td>
</tr>
<tr>
<td>hip front leg</td>
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<td>&lt;0.001*</td>
<td>0.361</td>
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<tr>
<td>hip rear leg</td>
<td>&lt;0.001*</td>
<td>&lt;0.001*</td>
<td>0.005*</td>
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Table A.4: Vertical weight distribution between the front and rear legs. Split squat step lengths of $l_1=55\%$ of leg length ($l_l$), $l_2=70\%$ $l_l$, and $l_3=85\%$ $l_l$ and four tibia angles ($a$) were evaluated. Significant differences in the vertical weight distribution between the legs for different step lengths and tibia angles are marked with a star [87].

<table>
<thead>
<tr>
<th></th>
<th>a 60°</th>
<th>a 75°</th>
<th>a 90°</th>
<th>a 105°</th>
</tr>
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<tbody>
<tr>
<td>$l_1$</td>
<td>0.66 ± 0.03</td>
<td>0.63 ± 0.03</td>
<td>0.57 ± 0.04</td>
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<tr>
<td>$l_2$</td>
<td>0.69 ± 0.02</td>
<td>0.64 ± 0.02</td>
<td>0.60 ± 0.03</td>
<td></td>
</tr>
<tr>
<td>$l_3$</td>
<td>0.70 ± 0.04</td>
<td>0.64 ± 0.03</td>
<td>0.60 ± 0.03</td>
<td>0.56 ± 0.03</td>
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</tbody>
</table>
Table A.5: Maximal normalized external moments ([Nm/kg], positive for external flexion) of the knee and hip joints during split squats with different step lengths. Split squat step lengths of \( l_1 = 55\% \) of leg length (LL), \( l_2 = 70\% \) LL, and \( l_3 = 85\% \) LL and four tibia angles (a) were evaluated. Significant differences in the moments of the joints during split squats with different step lengths and tibia angles are marked with a star. The maximal normalized moments of the knee and hip of the rear leg showed significant interactions of the tibia angles and step lengths. Therefore, a Bonferroni corrected pairwise comparison could not be performed [87].

<table>
<thead>
<tr>
<th></th>
<th>a 60°</th>
<th>a 75°</th>
<th>a 90°</th>
<th>a 105°</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>knee front leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 1</td>
<td>1.46 ± 0.14</td>
<td>1.38 ± 0.17</td>
<td>1.19 ± 0.22</td>
<td></td>
</tr>
<tr>
<td>1 2</td>
<td>1.40 ± 0.17</td>
<td>1.33 ± 0.20</td>
<td>1.14 ± 0.19</td>
<td></td>
</tr>
<tr>
<td>1 3</td>
<td>1.28 ± 0.12</td>
<td>1.19 ± 0.17</td>
<td>1.13 ± 0.21</td>
<td>0.80 ± 0.16</td>
</tr>
<tr>
<td><strong>knee rear leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 1</td>
<td>1.57 ± 0.15</td>
<td>1.68 ± 0.19</td>
<td>1.73 ± 0.22</td>
<td></td>
</tr>
<tr>
<td>1 2</td>
<td>1.45 ± 0.14</td>
<td>1.61 ± 0.17</td>
<td>1.71 ± 0.21</td>
<td></td>
</tr>
<tr>
<td>1 3</td>
<td>1.41 ± 0.17</td>
<td>1.57 ± 0.18</td>
<td>1.81 ± 0.20</td>
<td>1.82 ± 0.24</td>
</tr>
<tr>
<td><strong>hip front leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 1</td>
<td>1.72 ± 0.30</td>
<td>1.56 ± 0.30</td>
<td>1.42 ± 0.30</td>
<td></td>
</tr>
<tr>
<td>1 2</td>
<td>1.92 ± 0.20</td>
<td>1.69 ± 0.14</td>
<td>1.71 ± 0.32</td>
<td></td>
</tr>
<tr>
<td>1 3</td>
<td>2.27 ± 0.38</td>
<td>1.96 ± 0.31</td>
<td>1.82 ± 0.28</td>
<td>1.72 ± 0.29</td>
</tr>
<tr>
<td><strong>hip rear leg</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 1</td>
<td>-1.14 ± 0.23</td>
<td>-0.91 ± 0.19</td>
<td>-0.54 ± 0.17</td>
<td></td>
</tr>
<tr>
<td>1 2</td>
<td>-1.45 ± 0.30</td>
<td>-1.24 ± 0.25</td>
<td>-0.91 ± 0.23</td>
<td></td>
</tr>
<tr>
<td>1 3</td>
<td>-1.77 ± 0.23</td>
<td>-1.69 ± 0.14</td>
<td>-1.49 ± 0.31</td>
<td>-1.11 ± 0.27</td>
</tr>
</tbody>
</table>
A.2 Corresponding figures and tables of back extension exercises study (Chapter 4.2)

Figure A.5: Back Extension (BE) exercise on a 45° BE bench: a: subject, b: opto-electronic motion cameras, c: two decoupled force plates, d: reflective markers, e: EMG sensors, f: video camera [200].
Figure A.6: Moments (positive for external flexion moment) as a function of joint angle averaged over all repetitions and all subject plots are plotted for all BE exercise types (solid: BEh, dashed: BEs, blue: 2L, green: 1L, red: IB; *: starting point) a: Normalised knee moment in the sagittal plane [Nm/BW] as a function of the corresponding knee flexion angle [°]. b: Normalised hip moment in the sagittal plane [Nm/BW] as a function of the corresponding hip flexion angle [°]. c: Normalised back moment at the level L4/L5 in the sagittal plane [Nm/BW] as a function of the corresponding lumbar curvature [1/m]. d: Normalised thoracic curvature in the sagittal plane [1/m] [200].
Table A.6: The following instructions were given to the subject to ensure correct execution of the exercises [200].

<table>
<thead>
<tr>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Position yourself into the BE construction such that your upper body is in line with your legs.</td>
</tr>
<tr>
<td>2 Position both legs or only the left or the right leg into the construction according to the execution form that you are about to perform.</td>
</tr>
<tr>
<td>3 Make sure that your heels are positioned well on the platform of the lower part of the construction and that your knees are straight.</td>
</tr>
<tr>
<td>4 Slightly abduct your arms and rotate them externally.</td>
</tr>
<tr>
<td>5 Perform eight repetitions of the specific execution form:</td>
</tr>
<tr>
<td>6 For the BEh the lowest point is reached right before your spine starts to bend.</td>
</tr>
<tr>
<td>7 For the BEs the lowest point is reached right before your hip starts to flex.</td>
</tr>
<tr>
<td>8 The maximal extension is reached as soon as your upper body is in line with your legs.</td>
</tr>
<tr>
<td>9 Pay attention to the right breathing pattern during the execution.</td>
</tr>
</tbody>
</table>
Table A.7: Mean range of motion [°] of middle to upper back (RoMmub), lower to middle back (RoMlmb), pelvis to lower back (RoMplb), hip (RoMh) and knee (RoMk), lumbar and thoracic curvature [1/m] at starting (Cl,s, Ct,s) and reversal point (Cl,rp, Ct,rp) and maximal joint moments [Nm/BW] of back (Mb,max), hip (Mh,max) and knee (Mk,max) in the sagittal plane for the different types (BEh, BEs) and versions (1L, 2L and IB) of BE [200].

<table>
<thead>
<tr>
<th></th>
<th>RoMmub</th>
<th>RoMlmb</th>
<th>RoMplb</th>
<th>RoMh</th>
<th>RoMk</th>
<th>C_{l,s}</th>
<th>C_{l,rp}</th>
<th>C_{t,s}</th>
<th>C_{t,rp}</th>
<th>Mb,max</th>
<th>Mh,max</th>
<th>Mk,max</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>[°]</td>
<td>[°]</td>
<td>[°]</td>
<td>[°]</td>
<td>[°]</td>
<td>[1/m]</td>
<td>[1/m]</td>
<td>[1/m]</td>
<td>[1/m]</td>
<td>[Nm/kg]</td>
<td>[Nm/kg]</td>
<td>[Nm/kg]</td>
</tr>
<tr>
<td>1L</td>
<td>8.9±3.7</td>
<td>14.8±6</td>
<td>16.6±5</td>
<td>5.3±2.9</td>
<td>5.6±2.1</td>
<td>-0.1±1.1</td>
<td>2.4±0.7</td>
<td>1.8±0.8</td>
<td>1±0.2</td>
<td>1.1±0.2</td>
<td>-0.8±0.2</td>
<td></td>
</tr>
<tr>
<td>BEh</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2L</td>
<td>10±4.5</td>
<td>12.5±5.1</td>
<td>17.4±5</td>
<td>36.3±7.9</td>
<td>5.2±2</td>
<td>0±1.2</td>
<td>2.6±0.6</td>
<td>1.8±0.8</td>
<td>1.1±0.2</td>
<td>0.8±0.1</td>
<td>-0.5±0.1</td>
<td></td>
</tr>
<tr>
<td>IB</td>
<td>5.7±2.6</td>
<td>12.5±6.9</td>
<td>17.9±4.3</td>
<td>36±7.5</td>
<td>5.1±2.1</td>
<td>0.3±1.5</td>
<td>2.5±0.7</td>
<td>1.9±0.8</td>
<td>1±0.2</td>
<td>0.8±0.1</td>
<td>-0.5±0.1</td>
<td></td>
</tr>
<tr>
<td>1L</td>
<td>16.6±4.2</td>
<td>36±9.3</td>
<td>11.7±7.3</td>
<td>14.1±9.9</td>
<td>5.7±2.2</td>
<td>0±1.9</td>
<td>3.3±0.6</td>
<td>2±0.7</td>
<td>0.9±0.2</td>
<td>0.9±0.2</td>
<td>-0.6±0.2</td>
<td></td>
</tr>
<tr>
<td>BEs</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2L</td>
<td>14.7±3.2</td>
<td>35.9±10.1</td>
<td>13.8±8.6</td>
<td>15.5±11.7</td>
<td>5.4±2.1</td>
<td>-0.6±1.5</td>
<td>3.3±0.6</td>
<td>2.2±0.7</td>
<td>1±0.2</td>
<td>0.7±0.1</td>
<td>-0.4±0.1</td>
<td></td>
</tr>
<tr>
<td>IB</td>
<td>21.2±6</td>
<td>35.1±10.1</td>
<td>16.2±8.7</td>
<td>17.1±11.6</td>
<td>5.3±2.3</td>
<td>-0.6±1.5</td>
<td>3.4±0.6</td>
<td>2.2±0.8</td>
<td>0.9±0.2</td>
<td>0.7±0.1</td>
<td>-0.4±0.1</td>
<td></td>
</tr>
</tbody>
</table>

Appendix A
Appendix A

Table A.8: P-values for differences in mean range of motion, curvature and maximal joint moments in the sagittal plane within and between types and versions [200].

<table>
<thead>
<tr>
<th></th>
<th>RoMsub</th>
<th>RoMsub</th>
<th>RoMsub</th>
<th>RoMsub</th>
<th>Csub</th>
<th>Csub</th>
<th>Csub</th>
<th>Csub</th>
<th>Msubmax</th>
<th>Msubmax</th>
</tr>
</thead>
<tbody>
<tr>
<td>BEh: 1L ↔ 2L</td>
<td>0.744</td>
<td>0.413</td>
<td>1.000</td>
<td>0.002</td>
<td>0.000</td>
<td>0.080</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>0.000</td>
</tr>
<tr>
<td>BEh: 2L ↔ IB</td>
<td>0.001</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>1.000</td>
<td>0.470</td>
</tr>
<tr>
<td>BEs: 1L ↔ 2L</td>
<td>0.170</td>
<td>0.000</td>
<td>1.000</td>
<td>0.645</td>
<td>0.349</td>
<td>0.150</td>
<td>0.003</td>
<td>1.000</td>
<td>1.000</td>
<td>0.000</td>
</tr>
<tr>
<td>IB: BEh ↔ BEs</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>0.735</td>
<td>0.686</td>
<td>0.375</td>
<td>0.169</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
</tr>
<tr>
<td>IB: BEs ↔ BEh</td>
<td>0.000</td>
<td>0.000</td>
<td>0.327</td>
<td>0.302</td>
<td>0.308</td>
<td>0.012</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
<td>0.000</td>
</tr>
</tbody>
</table>

Table A.9: Mean curve parameters of M. gluteus maximus (GlutMax), M. gluteus medius (GlutMed), lateral hamstrings (HamLat), medial hamstrings (HamMed), lumbar part of M. erector spinae (ErecLum), thoracic part of M. erector spinae (ErecThor), M. rectus abdominis (Abdo) and M. obliquus abdominis (Obli) for the eccentric and concentric phase of BE [200].

<table>
<thead>
<tr>
<th></th>
<th>GlutMax</th>
<th>GlutMed</th>
<th>HamLat</th>
<th>HamMed</th>
<th>ErecLum</th>
<th>ErecThor</th>
<th>Abdo</th>
<th>Obli</th>
</tr>
</thead>
<tbody>
<tr>
<td>IB: BEh ↔ BEs</td>
<td>0.003</td>
<td>0.002</td>
<td>1.000</td>
<td>0.204</td>
<td>0.792</td>
<td>1.000</td>
<td>0.425</td>
<td>1.000</td>
</tr>
<tr>
<td>IB: BEs ↔ BEh</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.016</td>
<td>1.000</td>
<td>0.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>IB: BEh ↔ BEs</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.910</td>
<td>1.000</td>
<td>0.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>IB: BEs ↔ BEh</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.910</td>
<td>0.000</td>
<td>0.000</td>
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</tr>
</tbody>
</table>

Table A.10: P-values for muscle activity within and between types and versions [200].

<table>
<thead>
<tr>
<th></th>
<th>GlutMax</th>
<th>GlutMed</th>
<th>HamLat</th>
<th>HamMed</th>
<th>ErecLum</th>
<th>ErecThor</th>
<th>Abdo</th>
<th>Obli</th>
</tr>
</thead>
<tbody>
<tr>
<td>IB: BEh ↔ BEs</td>
<td>0.003</td>
<td>0.002</td>
<td>1.000</td>
<td>0.001</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.981</td>
</tr>
<tr>
<td>IB: BEs ↔ BEh</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.001</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>IB: BEh ↔ BEs</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.001</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>IB: BEs ↔ BEh</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>0.001</td>
<td>0.000</td>
<td>0.000</td>
<td>1.000</td>
<td>1.000</td>
</tr>
</tbody>
</table>

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Appendix B

Curriculum Vitae
General Information
Florian Schellenberg
MSc ETH in Human Movement Sciences with a Major in Biomechanics

ETH Zürich
Institute for Biomechanics
Leopold-Ruzicka-Weg 4
HCP H 16.3
8093 Zürich, Switzerland

Phone: +41 44 633 72 01
Mobile: +41 79 293 02 16
Email: fs@ethz.ch

Date of Birth: February 27, 1986
Place of Birth: Uster, ZH, Switzerland
Nationality: Swiss
### Educations and Positions

**since 2012** Ph.D. Student at the Institute for Biomechanics, ETH Zürich, Switzerland

**since 2013** Beach Soccer Coach from Winti Panthers at Winterthur, Switzerland

**since 2013** Beach Soccer Coach from ASVZ at Zürich, Switzerland

**since 2011** Kids Soccer Coach at Zürich, Switzerland

- 2013 Foundation of **LLS Biomechanics GmbH** company.
- 2013 3 month **Stay Abroad** in Central and South America, inclusive Spanish class
- 2012 **Scientific Assistant** at the Institute for Biomechanics, ETH Zürich, Switzerland
- 2009 - 2012 **Master of Science** in Biomechanics, ETH Zürich, Switzerland
    - Development of a new devices to determine the bone strength intra-operatively
    - Project management
    - Measurement at artificial bone
    - Evaluation using Matlab and SPSS
  - 3 month internship in the field of human movement sciences, Universitätspital Zürich, Switzerland
    - Classification of movement pattern with accelerometer on hemiparetic patients
    - Development of the measurement set up
    - Patient recruitment and measurements
    - Evaluation and movement classification using Matlab
    - Literature review
  - 6 month internship à 50% in the scientific field of gait analysis, ETH Zürich, Switzerland
    - Gait analysis with different shoes
    - Kinetic, kinematic and EMG measurements using Nexus
    - Patient recruitment and measurements
    - Analyses and evaluation using Matlab
    - Development of upper joint angle determination and EMG processing
- **Graduation:** February, 2012

**2009-2012** **Tutor** in Biomechanics, ETH Zürich, Switzerland
- Teaching students at Bachelor (5th, 6th semester) and Masters Degree (7th semester)

**2011** **Ski Instructor** in Churwalden, Switzerland
- Group and private lessons to children and adults

**2006 - 2010** **Bachelor of Science** in Human Movement Sciences and Sport
- **Graduation:** October, 2010

**2006** 3 month **Stay Abroad** in South Africa

**2005 - 2006** **Storeman**, Balcar AG, Uster, Switzerland
1999 - 2005 Eidgenössische (Federal) Matura, Kantonsschule Zürcher Oberland, Switzerland
• mathematical and natural science profile

Languages
- German nativ
- English advanced (C1)
- French basic knowledge (B1)
- Spanish basic knowledge (A2)

IT - Knowledge
- MS-Office
- MATLAB
- LaTeX
- OpenSim SimTK
- Mathematica
- Java, Phython, C, PHP, HTML, SPSS

Other Activities
Beachsoccer Coach and Player, Snowsport, Soccer Coach

Advisor
1. Ramona Häberle, 2016, HEST, ETH Zürich, Switzerland, "Comparison of the kinematics and kinetics of shoulder exercises performed with constant and elastic resistance.", Master Thesis
2. Mira Ostermann, 2016, Bern University of Applied Sciences, Switzerland, "Influences on Knee Alignment during Squats in various Standce Widths and Foot Positions.", Master Thesis
3. Edeny Baaklini, 2016, Bern University of Applied Sciences, Switzerland, "Cross-sectional study on the influence of the height of highheeled shoes on the kinematics and kinetics of the pelvis and the spine during walking.", Master Thesis
4. Sabrina Windmüller, 2016, University of Freiburg; Hochschule fur Sport, Magglingen, Switzerland, "Welche Imitationssprünge auf dem Messwagen sind im Kraft-Zeit-Verlauf denen auf der Skisprungschanze am ähnlichsten?", Master Thesis
5. Fabian Zeidel, 2016, FH Technikum Wien, Austria, "Einfluss der Standbreite und der Fussposition auf die auftretenden Momente im Kniegelenk, Hüftgelenk und in der Lendenwirbelsäule bei Kniebeugen.", 3 month Internship
6. Pia Magdalena Zimmer, 2016, FH Technikum Wien, Austria, "Influence of Stance Width and Foot Position on Sagittal Curvature of the Lumbar Spine during Back Squats.", 3 month Internship
7. Santiago Alzate Restrepo, 2016, Profesional en Entrenamiento Deportivo, Universidad de Antioquia, Medellín, Colombia, “Biomechanical analysis procedures in sports environment.”, 2 month Internship
8. Srivatsan Yadhunathan, 2016, MATH, ETH Zürich, Switzerland, "Different cost functions using static optimizations and statistical analyses of muscle forces during different strength exercises.", 3 month Internship
9. Nicole Schmid, 2015, HEST, ETH Zürich, Switzerland, "EMG Data Processing and Sensitivity Analysis in Data inputs within an OpenSim Process.", 3 month Internship
11. Frédéric Lamon, 2015, HEST, ETH Zürich, Switzerland, "Comparison between Foot-Back and Foot-Up stance serve in tennis in relation to performance, lower limbs loading conditions and muscles force development.", Master Thesis
12. Thomas Lamparter, 2015, EHS Magglingen, Switzerland, "Validität und Reliabilität von einfachen Messgeräten zur Bestimmung der Langhantelgeschwindigkeit bei freien Kniebeugen.", Master Thesis
13. Michael Plüss, 2015, HEST, ETH Zürich, Switzerland, "Modellierung von Krafttrainingsübungen.", 3 month Internship
14. Anna Kratschmar, 2015, Universität zu Köln, D, "Biomechanics of the strength exercise Back Extension.", 2 Month Internship
15. Christian Grether, 2015, MAVT, ETH Zurich, "Loading Conditions and Simulation of Muscle Forces during Squats.", Semester Project
16. Ramona Häberle, 2015, HEST, ETH Zurich, "Biomechanics of the strength exercise Back Extension.", 3 Month Internship
17. Nicole Hörterer, 2015, HEST, ETH Zurich, "EMG data processing of the Strength Exercise Back Extension.", 3 Month Internship
18. Mazzariello Michele, 2014, HEST, ETH Zurich, "Biomechanics of the strength exercise Back Extension.", Semester Project
19. Edeny Baaklini, 2014, HEST, ETH Zurich, "Back Extension.", Semester Project
20. Janine Weigand, 2014, Fachhochschule Koblenz, RheinAhrCampus, Remagen, D, "Design of the training equipment and biomechanical analyses of the strength exercise back extension.", 3 Month Internship
22. Stefan Anthamatten, 2013, MAVT, ETH Zurich, "Feasibility Study: Influence of damping elements on acceleration and motion on racket and lower arm during tennis serves", Bachelor Thesis
23. Olivier Meyer, 2013, HEST, ETH Zurich, "Generation and analysis of Test Cases using in the Institute for Biomechanics", 3 Month Internship

Publications

Thesis
Public Articles

Presentations & Conference Proceedings
5. Angst, M., et al., Load condition of the wrist during the forward handspring, the forward handspring with ulnar deviated hand positioning and the backward handspring. Presentation at ISBS as co-author, Poitiers, France, 2015.
13. Angst, M., et al., Load condition of the wrist during the forward handspring, the forward handspring with ulnar deviated hand positioning and the backward handspring. Poster presentation at SGS as co-author, Basel, Switzerland, 2013.

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**Patents**


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**Research Society Memberships**

- since 2015: International Society of Biomechanics in Sports
- since 2013: Sportwissenschaftliche Gesellschaft der Schweiz

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Zuerich, 17. August 2016