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A comparison of optimization methods and knee joint degrees of freedom on muscle force predictions during single-leg hop landings

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Abstract

The aim of this paper was to compare the effect of different optimization methods and different knee joint degrees of freedom (DOF) on muscle force predictions during a single legged hop. Nineteen subjects performed single-legged hopping manoeuvres and subject-specific musculoskeletal models were developed to predict muscle forces during the movement. Muscle forces were predicted using static optimization (SO) and computed muscle control (CMC) methods using either 1 or 3 DOF knee joint models. All sagittal and transverse plane joint angles calculated using inverse kinematics or CMC in a 1 DOF or 3 DOF knee were well-matched (RMS error < 3^{o}). Biarticular muscles (hamstrings, rectus femoris and gastrocnemius) showed more differences in muscle force profiles when comparing between the different muscle prediction approaches where these muscles showed larger time delays for many of the comparisons. The muscle force magnitudes of vasti, gluteus maximus and gluteus medius were not greatly influenced by the choice of muscle force prediction method with low normalized root mean squared errors (< 48%) observed in most comparisons. We conclude that SO and CMC can be used to predict lower-limb muscle co-contraction during hopping movements. However, care must be taken in interpreting the magnitude of force predicted in the biarticular muscles and the soleus, especially when using a 1 DOF knee. Despite this limitation, given that SO is a more robust and computationally efficient method for predicting muscle forces than CMC, we suggest that SO can be used in conjunction with musculoskeletal models that have a 1 or 3 DOF knee joint to study the relative differences and the role of muscles during hopping activities in future studies.

Keywords: Static Optimization, Computed Muscle Control, Musculoskeletal model, hopping, muscle co-contraction, knee joint

Introduction

Accurate knowledge of lower-limb muscle forces is important in understanding how muscles function during normal and pathological gait. Reliable estimations of muscle forces can improve predictions of joint contact forces and stresses (Kim et al., 2009) as well as ligament forces (Kernozek and Ragan, 2008; Laughlin et al., 2011; Mokhtarzadeh et al., 2013). A collective understanding of these biomechanical variables can provide insight into the causes or consequences of different joint diseases. For example, accurate knowledge of knee muscle forces can be utilized to improve our understanding of changes in medial and lateral tibiofemoral contact forces after an anterior cruciate ligament injury, which has been suggested to be precursor to knee osteoarthritis (Fregly et al., 2012).

Musculoskeletal modelling has recently become a powerful biomechanical tool used to predict muscle forces in which optimization methods are commonly utilized to solve the muscle-moment redundancy problem (i.e. a net joint moment can be produced from an infinite number of muscle force combinations) (Crowninshield, 1978). Static optimization (SO) and computed muscle control (CMC) are two popular optimization methods used for predicting muscle forces and are accessible for use in the freely available musculoskeletal modelling software, OpenSim (Delp et al., 2007; Thelen and Anderson, 2006). SO is an inverse dynamics-based method that partitions the net joint moment amongst individual muscles by minimizing a given performance criterion (e.g. sum of squares of muscle activations) (Erdemir et al., 2007). On the other hand, CMC is a forward dynamics-based approach that utilizes feedback control theory to predict a set of muscle excitations that will produce kinematics that closely match the kinematics calculated from inverse kinematics (Thelen and Anderson, 2006; Thelen et al., 2003). Whilst these methods provide a means for obtaining otherwise unattainable *in vivo* muscle forces, these predictions are limited in that it is challenging to know how valid or accurate these methods are in predicting individual muscle forces given that no direct measures are available.

A previous study has shown that the muscle forces predicted by SO can produce accurate joint contact forces during walking by comparing the predicted contact forces to those measured in a person with an instrumented knee implant (Kim et al., 2009). Previous studies have also shown that SO and CMC produce similar muscle force predictions during walking and running in terms of timing and magnitude (Anderson and Pandy, 2001a; Lin et al., 2011). However, these studies have cautioned against the use of SO for ballistic movements such as jumping as SO may produce muscle activation patterns that are inconsistent with electromyographic (EMG) recordings (Lin et al., 2011). In addition, the ability of SO to predict co-contraction of antagonistic muscles has been criticised because this method excludes muscle activation dynamics. However, several studies have mathematically proven that multi-jointed models containing joints with multiple degrees of freedom (i.e. non-planar joints) can predict co-contraction of antagonistic muscles (Ait-Haddou et al., 2000; Jinha et al., 2006a, 2006b). Given that many past studies have used planar knee joint models i.e. 1 degree of freedom (DOF) when predicting muscle forces (Dorn et al., 2012; Fok et al., 2013; Mokhtarzadeh et al., 2013), the current study aims to evaluate the forces generated by the lower-limb muscles using different optimization methods and knee degrees of freedom.

Therefore, our study proposes to compare the individual lower-limb muscle force results produced by SO and CMC using both planar and non-planar knee joint models during a ballistic movement (i.e., hopping). We hypothesise that the muscle force results based on the SO method using a 3 degree-of-freedom (DOF) knee joint will be similar to those based on the CMC method from both a 1 and 3 DOF knee joint (H_1). On the other hand, we estimate that SO results from a 1 DOF knee joint will be significantly lower than the results obtained from other combinations of knee joint types and optimization methods (H_2).

Methods

Nineteen healthy and physically active subjects with no history of knee injury (height = 1.74 ± 0.08 m, body mass = 74.2 ± 10.8kg) participated in this study after providing informed consent. Ethical approval was provided by the University of Melbourne's Behavioural and Social Sciences Human Ethics sub-committee (ethics ID 1136167). Data were collected in the Physiotherapy Movement Laboratory at The University of Melbourne.

Participants performed an initial static trial by standing in a neutral position and subsequently completed multiple trials of a single leg hop task. On average two trials per subject were simulated in this study. The distanced hopped was standardised to the participant's leg length and upper limb movement was standardised by asking participants to fold their arms across their chest. Small reflective markers were mounted on the trunk and both lower limbs of participants. Marker trajectories and ground reaction forces (GRF) were collected simultaneously using a 14 camera Vicon motion analysis system and ground-embedded AMTI force plates. Ground reaction force and marker data were collected at 2400 Hz and 120 Hz respectively. Electromyographic (EMG) activity was collected simultaneously with an eight channel Noraxon EMG system (Noraxon USA Inc., Scottsdale, Arizona) sampling at 2400 Hz using non-preamplified skin mounted Ag/Cl electrodes (Duotrode, Myotronics). EMG data were collected from the vastus lateralis, vastus medialis, rectus femoris, lateral and medial hamstrings, gluteus medius and medial gastrocnemius muscles of the subject's dominant leg. A similar filtering method applied in previous studies was utilized for the EMG data (Laughlin et al., 2011; Mokhtarzadeh et al., 2013).

All musculoskeletal modelling and analyses were performed using OpenSim (Delp et al., 2007), MATLAB and the Edward cluster, a high performance computing (HPC) service, at The University of Melbourne. Kinematic and kinetics data were filtered using butterworth filter with a 4th order, zero-lag, recursive filter with a cut-off frequency of 15 Hz.

Two different subject-specific musculoskeletal models were generated for each participant (1 DOF and 3 DOF knee) by scaling generic models according to body segment dimensions recorded from the static trial. Both models consisted of 92 musculotendon units i.e. Gait2392 model in OpenSim. The model with 1 DOF knee had 23 degrees-of freedom while the model with 3 DOF knee had 27 degrees-of-freedom. Musculotendon units were modelled as a three element Hill-type model (Zajac, 1989). The ankle was modelled as 1 DOF joint whereas hip joint consisted of 3 DOF. The maximum isometric force property of each muscle was scaled by a factor of 3 to account for differences in muscle strength between our healthy young adults and the cadavers, which our generic models are based on (Dorn et al., 2012). For each trial and for both models, inverse kinematic analyses were used to calculate joint kinematics by minimizing the distance between model and measured marker trajectories (Lu and O'Connor, 1999) while joint moments were calculated using a traditional inverse dynamics approach. Two optimization methods (SO and CMC) were implemented separately to predict muscle forces to give a total of four approaches for muscle force prediction: (i) SO with 1 DOF knee, (ii) SO with 3 DOF knee, (iii) CMC with 1 DOF knee, and (iv) CMC with 3 DOF knee. SO partitions the net joint moments into individual muscle forces by minimising the sum of muscle activations squared at each time instant of the hop-landing cycle (Anderson and Pandy, 2001b). CMC performs a forward simulation to compute a set of muscle excitations that will drive the model to track the experimentally-derived joint angular accelerations. Tracking of joint kinematics is achieved through using a proportional-derivative controller while the required set of muscle excitations are calculated using SO (Thelen and Anderson, 2006; Thelen et al., 2003).

All analyses were performed over the eccentric landing phase of the task, which encompassed the period from initial foot strike to maximum knee flexion (Mokhtarzadeh et al., 2013). Foot strike was defined as the moment at which vertical GRF just reached above a predefined force (i.e., >10N) and then CMC and SO results were synchronized to account for the time CMC requires to initialize. Landing phase was defined from the time CMC and SO were synchronized to maximum knee flexion angle (0-100%). Using musculoskeletal modelling, nine major lower-limb muscles were compared including vasti (VAS), rectus femoris (RF), gluteus maximus (GMAX), gluteus medius (GMED), hamstrings (HAMS), gastrocnemius (GAS), and soleus (SOL). GMAX and SOL comparisons did not involve EMG. A cross-correlation was performed to compare the similarity in the shape of each muscle force time profile for the four different muscle force prediction approaches. This analysis calculated the time delay required to achieve the maximum unbiased correlation coefficient (R). Specifically, the unbiased correlation coefficient and time delay were calculated by displacing the muscle force profile in time predicted by one method relative to another method (from -100% to 100% of the landing phase) and subsequently, taking the maximum value for the correlation at the time displacement required to achieve this maximum value. For each trial the cross-correlation was performed between the signals resulting from two different methods. The cross-correlation results are a measure of correlation and a measure of time displacement (positive time displacement indicates the first profile has a delay over the second profile, whereas negative means that the first profile has a lead over the second profile). For each comparison, the mean and standard deviation across all trials were calculated for the unbiased correlation coefficient and time displacement required (hereafter called the time delay).

For each muscle, a normalized root mean squared error (NRMSE) was also calculated between the time-shifted muscle force profiles to compare differences in the magnitude of muscle force predictions. For each muscle, the NRMSE was normalized by the mean force over the entire landing phase and over all muscle force prediction approaches.

Results

All sagittal and transverse plane joint angles calculated using inverse kinematics or CMC in a 1 DOF or 3 DOF knee were well-matched (RMS error < 3^{o}) (Figure 1). Residual moments and forces across all participants were also within an acceptable range (RMS < 0.2 BW for residual forces and RMS < 0.05 BW-HT for residual moments) (Figure 2). Finally, muscle force profiles were qualitatively consistent with EMG measurements using all four muscle force prediction approaches (Figure 3).

Muscle Force Time History Profile

The time history profiles were similar for the GMED, VAS and SOL for all muscle force prediction approaches where most comparisons resulted in high correlation coefficients (R > 0.7) and time delays of less than 7.5% of the landing phase cycle (Table 1) (Figure 3). The force profiles of GMAX were similar for comparisons which did not involve the combination of CMC with a 3 DOF knee (R > 0.7 and time delays < 8% of the landing phase cycle).

Biarticular muscles (HAMS, RF and GAS) showed more differences in muscle force profiles when comparing between the different muscle prediction approaches where these muscles showed larger time delays for many of the comparisons (time shift > 8% landing phase) and moderate correlations (0.5 < R< 0.7) for the majority of the comparisons (Table 1).

The time profile of SOL was influenced by the choice of knee joint used as small time delays (<3% landing phase) were observed when comparing between the results that used the same knee joint model (Table 1). However, when comparing the muscle force profiles between different knee joint models, large time delays were noticed (>9% landing phase) (Table 1) (Figure 3).

Muscle Force Magnitude

In general, the muscle force magnitudes of VAS, GMAX and GMED were not greatly influenced by the choice of muscle force prediction method with low NRMSE (< 48%) observed in most comparisons (Table 2) (Figure 3). Bi-articular muscles, HAMS, RF and GAS, generally showed the greatest difference in magnitude between muscle force prediction methods (NRMSE > 112 % for most comparisons). Specifically, the combination of CMC and a 3 DOF knee produced considerably higher HAMS (NRMSE > 163%), GAS (NRMSE > 128%) and RF (NRMSE > 112% BW) forces than in the other optimization-knee joint combinations (Table 3). Similarly large differences in magnitude (NRMSE > 75%) were also observed in SOL where the use of a 3 DOF knee joint resulted in considerably lower SOL force than in a 1 DOF knee (Table 2) (Figure 3). The most similar muscle force magnitude predictions were seen when comparing the predictions from the SO method with a 1 DOF knee and CMC with a 1 DOF knee (NRMSE < 39%).

Discussion

The aim of this study was to compare the muscle force predictions given from two different optimization methods (SO and CMC) during a single-leg hopping movement in musculoskeletal models with planar knee joints and models with non-planar knee joints. In general, all four approaches predicted similar muscle force time histories/profiles. However, the magnitude of muscle forces predicted by CMC tended to be higher than SO in most of the major muscles for a given type of knee joint. Also, the use of a 3 DOF knee joint tended to result in larger muscle force predictions than a 1 DOF knee joint when assessing each optimization method independently. However, soleus was an exception to the abovementioned cases as CMC produced less force in soleus than SO for a particular type of knee joint. Furthermore, for a particular optimization method, the use of a 3 DOF knee joint resulted in less force in the soleus than the use of 1DOF knee joint.

The results of our study suggest that SO can predict less force output for knee-spanning muscles (HAMS, RF and GAS) when using a 1 DOF knee joint. The reasons for this could be primarily twofold: (1) the SO solution neglects excitation-activation dynamics and, (2) the SO solution only needs to find a combination of muscle forces to satisfy the knee kinematics in one plane (sagittal). However, when more DOFs are included in the knee, the optimization solution must find a combination of muscle forces to match knee kinematics in all three planes. Consequently, greater forces and different muscle force activation patterns may be needed from all knee-spanning muscles to closely match kinematics in all three planes. Nonetheless, it is important to not dismiss SO's ability to predict co-contraction in a 1 DOF knee joint as it still predicted muscle co-contraction, albeit at a lower magnitude. In addition, given that greater co-contraction of knee-spanning is occasionally assumed to be related to greater knee stiffness in clinical practice (Erdemir et al., 2007), one must be careful with making conclusions about knee stiffness during ballistic movements since the magnitude of muscle force predictions are influenced by the choice of knee joint and the type of optimization method used.

Interestingly, the force in the soleus was substantially lower when using a 3 DOF knee despite it being a uni-articular muscle. It seems that the greater co-contraction of knee spanning muscles predicted when using a 3 DOF knee joint corresponded with a redistribution of the ankle plantarflexor moment from the soleus to the gastrocnemius where there was a substantial decrease in soleus force and a substantial increase in gastrocnemius force. Hence, future studies involving the prediction of ankle plantarflexor muscle forces should carefully consider the choice of knee joint to be used as it will greatly influence the magnitude of forces predicted in these muscles. Furthermore, this finding has implications for the conclusions drawn from previous studies that have used SO to predict ankle muscle forces. For example, one study suggested that SOL has a role in protecting the ACL and based this deduction from the HAMS-to-SOL force ratio and the contribution of the SOL and GAS to the ACL force during single-leg landing (Mokhtarzadeh et al., 2013) whilst another study calculated the contribution of SOL and GAS to the centre of mass acceleration during running at different speeds (Dorn et al., 2012). It is possible that conclusions drawn from these studies could be different if they had used a 3 DOF knee joint (rather than a 1 DOF knee) in their analysis.

Interestingly, when SO was used in conjunction with a non-plantar knee joint, the magnitude of muscle force predictions were generally similar to that predicted by CMC in most cases regardless of the type of knee joint. Furthermore, all combinations of optimization methods and types of knee joints produced similar muscle force profiles for the major muscles in terms of their general shape. Given that SO is more computationally efficient (approximately five times more efficient) than CMC (Lin et al., 2011), has less preparation time and is more robust than CMC, it seems questionable whether there are justifiable benefits in including activation dynamics as a means of improving muscle force predictions during single-leg hopping. Our study suggests that the use of SO may provide an efficient alternative to CMC whilst yielding similar results - particularly for uni-articular muscles.

This study builds upon previous findings showing that similar muscle forces can be predicted for dynamic optimization and SO during walking (Anderson and Pandy, 2001a) and for CMC and SO during walking and running (Lin et al., 2011) by extending the analysis to a more ballistic type of movement (e.g. single-leg hopping). Unlike the current study, these previous studies only used musculoskeletal models with a 1 DOF knee and incorporated tasks that are more cyclic and less physically demanding, which may not be greatly influenced by muscle activation dynamics. Nonetheless, our results were similar to these previous studies in that for a chosen type of knee joint, SO and CMC generally produced similar muscle force time profiles during single-leg hopping (Figure 3). Furthermore, it should be noted that the conclusions from previous studies were founded upon a single trial from one subject whilst our study's findings were based on multiple trials from multiple subjects, which give us confidence in the conclusions we have deduced.

While musculoskeletal models provide a great tool in studying otherwise unattainable muscle forces, this approach does come with limitations. Firstly, it is impossible to know which optimization method and knee joint combination produced the most accurate muscle forces given it is extremely difficult and invasive to measure muscle force *in vivo*. It is possible that the magnitude and timing of all muscle force predictions are incorrect. In addition, our results for 3DOF models could have been influenced by off-plane (transverse and frontal) kinematic errors (Li et al., 2012). Nonetheless, EMG measurements were qualitatively consistent with muscle force time profiles predicted using all four muscle force prediction approaches so that we at least have confidence in the timing of our muscle force predictions (Figure 3). Furthermore, even if the magnitude of muscle forces were inaccurate, we have confidence in the validity of our comparisons given the kinematics were well-matched (Figure 1) and residual forces and moments were small (Figure 2). Secondly, the conclusions obtained from our study apply to single-leg hopping in healthy adults. It is unclear whether the same conclusions can be extended to other ballistic movements such as cutting and jumping.

In light of our findings and those of earlier studies, we conclude that both SO and CMC can be used to predict lower-limb muscle co-contraction during hopping movements. However, care must be taken in interpreting the magnitude of force predicted in the biarticular muscles and the soleus, especially when using a 1 DOF knee. Despite this limitation, given that SO is a more robust and computationally efficient method for predicting muscle forces than CMC, we suggest that SO be used in conjunction with musculoskeletal models that have a 1 DOF or 3 DOF knee joint to study the relative differences and role of muscles during hopping activities in future studies,; however, there is no agreement on which optimization method can better predict muscle forces during hopping.

Conflict of interest statement

None of the authors above has any financial or personal relationship with other people or organizations that could inappropriately influence this work, including employment, consultancies, stock ownership, honoraria, paid expert testimony, patent applications/registrations, and grants or other funding.

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Captions:

Figure 1: Joint angles during single-leg hopping calculated from computed muscle control (CMC, solid lines) and inverse kinematics (dashed lines) when using a 1 degree-of-freedom knee joint (in black) and 3 degree-of-freedom knee joint (in grey).

Figure 2: Residual moments and forces during single-leg hopping calculated from computed muscle control (CMC, solid lines) and static optimization (SO, dashed lines) when using a 1 degree-of-freedom knee joint (in black) and 3 degree-of-freedom knee joint (in grey).

Figure 3: Muscle forces during single-leg hopping predicted from computed muscle control (CMC) in a 1 degree of freedom (DOF) knee joint (solid black line) and in a 3 DOF knee joint (solid grey line), static optimization (SO) in a 1 DOF knee joint (dashed black line) and 3 DOF knee joint (dashed grey line). Muscle EMG (mean ± std) is shown as shaded regions. BW; body weight

Table 1. Cross-Correlation Results (Correlation Coefficient and Time Delay) for different muscles to compare SO, CMC and knee degrees of freedoms.

Table 2. Magnitude differences (Normalized root mean squared error) for different muscles to compare SO, CMC and knee degrees of freedoms.

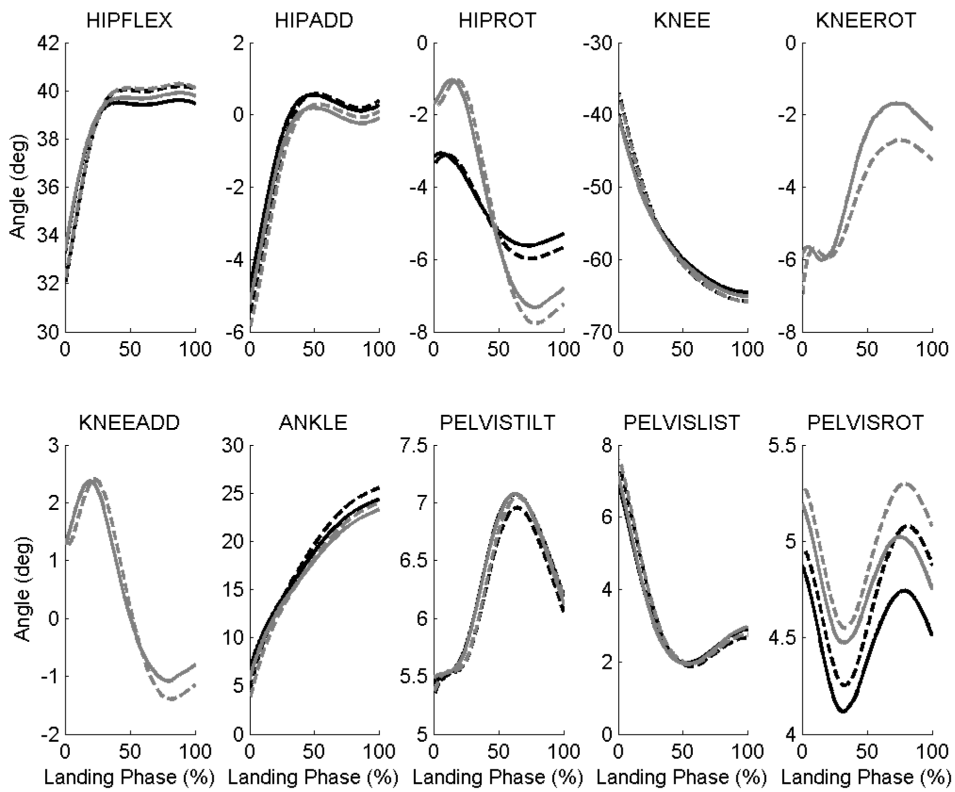
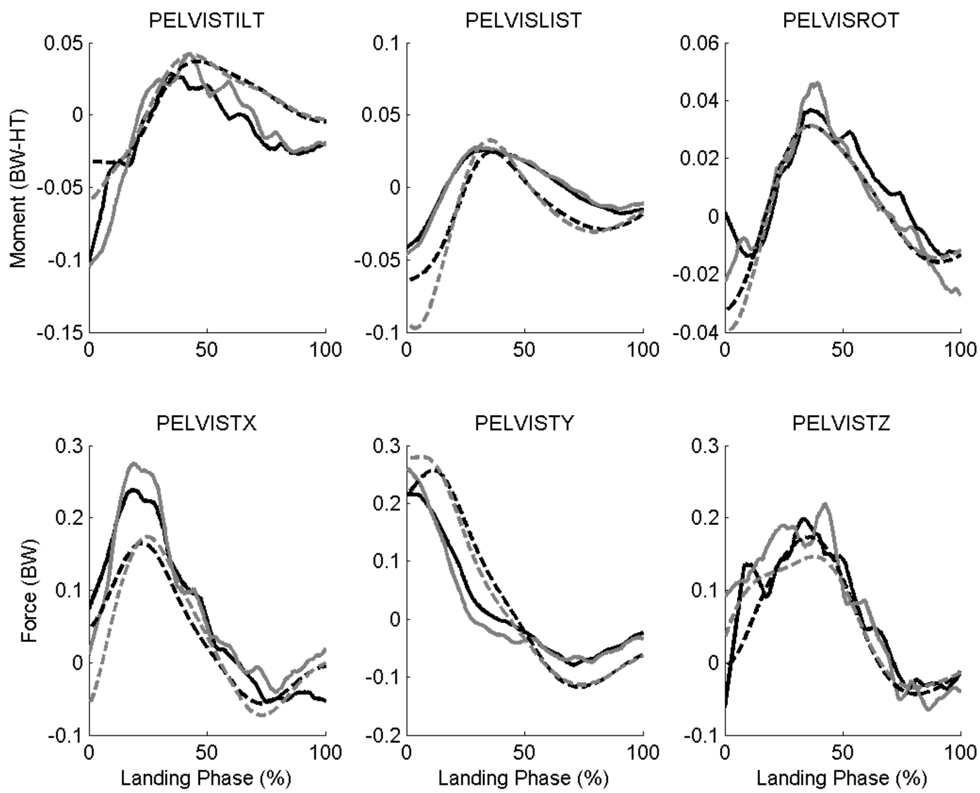


Figure 1: Joint angles during single-leg hopping calculated from computed muscle control (CMC, solid lines) and inverse kinematics (dashed lines) when using a 1 degree-of-freedom knee joint (in black) and 3 degree-of-freedom knee joint (in grey).



402

403 **Figure 2:** Residual moments and forces during single-leg hopping calculated from computed
404 muscle control (CMC, solid lines) and static optimization (SO, dashed lines) when using a 1
405 degree-of-freedom knee joint (in black) and 3 degree-of-freedom knee joint (in grey**).** The top
406 graphs present Pelvic rotations including tilt, list and rotation), and the bottom graphs show
407 pelvic translations i.e. anteroposterior, vertical and mediolateral respectively.

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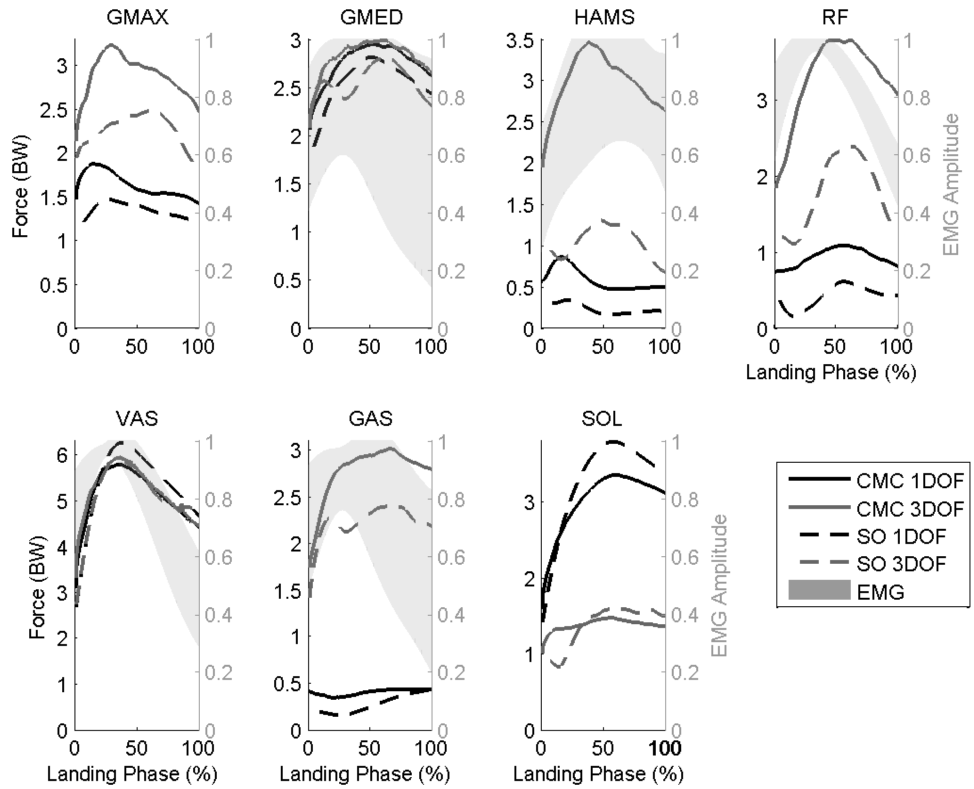

Figure 3: Muscle forces during single-leg hopping predicted from computed muscle control (CMC) in a 1 degree of freedom (DOF) knee joint (solid black line) and in a 3 DOF knee joint (solid grey line), static optimization (SO) in a 1 DOF knee joint (dashed black line) and 3 DOF knee joint (dashed grey line). Muscle EMG (mean ± std) is shown as shaded regions. BW; body weight

Table 1. Cross-Correlation Results (Correlation Coefficient and Time Delay) for different muscles to compare SO, CMC and knee degrees of freedoms.

		GMAX	GMED	HAMS	RF	VAS	GAS	SOL
Correlation Coefficient (R)								
SO 1DOF vs. SO 3DOF	mean	0.77	0.75	0.59	0.68	0.88	0.51	0.72
	std	0.19	0.16	0.18	0.20	0.15	0.15	0.19
SO 1DOF vs. CMC 1DOF	mean	0.85	0.89	0.78	0.80	0.93	0.65	0.92
	std	0.11	0.10	0.19	0.15	0.06	0.17	0.13
SO 1DOF vs. CMC 3DOF	mean	0.63	0.77	0.55	0.66	0.80	0.64	0.71
	std	0.18	0.16	0.16	0.19	0.16	0.19	0.19
SO 3DOF vs. CMC 1DOF	mean	0.77	0.76	0.61	0.63	0.81	0.56	0.74
	std	0.16	0.16	0.19	0.17	0.14	0.18	0.19
SO 3DOF vs. CMC 3DOF	mean	0.70	0.75	0.57	0.67	0.77	0.69	0.65
	std	0.16	0.17	0.18	0.18	0.19	0.18	0.19
CMC 1DOF vs. CMC 3DOF	mean	0.66	0.81	0.55	0.64	0.80	0.62	0.75
	std	0.20	0.14	0.14	0.19	0.15	0.18	0.18
Time delays (% landing phase)								
		GMAX	GMED	HAMS	RF	VAS	GAS	SOL
SO 1DOF vs. SO 3DOF	mean	5.5	3.0	11.9	2.8	-2.9	-11.2	7.8
	std	21.6	7.7	40.9	21.3	13.2	35.9	23.7
SO 1DOF vs. CMC 1DOF	mean	-1.2	0.1	10.4	-4.7	-0.2	-12.9	1.3
	std	2.6	5.4	28.6	13.2	0.9	22.4	6.7
SO 1DOF vs. CMC 3DOF	mean	-4.1	5.6	17.9	-5.2	-2.4	-3.8	5.8
	std	26.3	23.1	34.0	20.3	6.2	37.7	31.2
SO 3DOF vs. CMC 1DOF	mean	-7.0	1.4	-5.6	-1.9	3.5	-12.9	-4.1
	std	24.0	17.6	41.9	23.6	10.5	33.2	17.0
SO 3DOF vs. CMC 3DOF	mean	0.9	2.1	3.1	-8.3	-0.8	-3.9	-5.0
	std	15.1	17.3	31.2	23.3	15.1	21.9	19.7
CMC 1DOF vs. CMC 3DOF	mean	0.2	2.1	6.8	-10.5	-1.0	5.1	2.6
	std	23.5	14.5	35.6	33.2	4.8	38.2	19.1

*grey highlights: R > 0.7 or time delay < 7.5% landing phase. Std stands for Standard deviation. The grey highlights represents when the mean correlation coefficient (R) is greater than 0.7 or when the time delay is less than 7.5% of the landing phase. Negative values denote that the first listed method in the comparison best matches the second listed method, when the muscle force time curve is shifted by the reported value. For example, in a SO 1DOF vs. CMC 1DOF comparison for GMAX, a time shift value of -1.2 means that the muscle force time curve predicted using SO 1DOF needs to be shifted 1.2% earlier in the landing phase to produce a correlation coefficient of 0.85.

Table 2. Magnitude differences (Normalized root mean squared error) for different muscles to compare SO, CMC and knee degrees of freedoms.

		GMAX	GMED	HAMS	RF	VAS	GAS	SOL
SO 1DOF vs. SO 3DOF	mean	48%	14%	74%	91%	16%	134%	98%
	std	28%	5%	31%	48%	15%	39%	46%
SO 1DOF vs. CMC 1DOF	mean	18%	10%	35%	38%	10%	15%	21%
	std	5%	3%	23%	21%	5%	9%	18%
SO 1DOF vs. CMC 3DOF	mean	77%	15%	234%	186%	26%	180%	92%
	std	32%	7%	72%	49%	14%	49%	22%
SO 3DOF vs. CMC 1DOF	mean	38%	18%	55%	72%	19%	129%	87%
	std	26%	4%	32%	46%	13%	41%	42%
SO 3DOF vs. CMC 3DOF	mean	41%	18%	163%	112%	19%	56%	38%
	std	20%	8%	78%	41%	9%	52%	30%
CMC 1DOF vs. CMC 3DOF	mean	65%	12%	193%	140%	25%	170%	76%
	std	33%	6%	64%	53%	12%	43%	20%

*grey highlights: normalized root mean squared error < 75%. Std stands for Standard deviation.