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The effect of accounting for biarticularity in hip flexor and hip extensor joint torque representations

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ABSTRACT

Subject-specific torque-driven models have ignored biarticular effects at the hip. The aim of this study was to establish the contribution of monoarticular hip flexors and hip extensors to total hip flexor and total hip extensor joint torques for an individual and to investigate whether torque-driven simulation models should consider incorporating biarticular effects at the hip joint. Maximum voluntary isometric and isovelocity hip flexion and hip extension joint torques were measured for a single participant together with surface electromyography. Single-joint and two-joint representations were fitted to the collected torque data and used to determine the maximum voluntary joint torque capacity. When comparing two-joint and single-joint representations, the single-joint representation had the capacity to produce larger maximum voluntary hip flexion torque (larger by around 9% of maximum torque) and smaller maximum voluntary hip extension torque (smaller by around 33% of maximum torque) with the knee extended. Considering the range of kinematics found for jumping movements, the single-joint hip flexors had the capacity to produce around 10% additional torque, while the single joint hip extensors had about 70% of the capacity of the two-joint representation. Two-joint representations may overcome an over-simplification of single-joint representations by accounting for biarticular effects, while building on the strength of determining subject-specific parameters from measurements on the participant.

1. Introduction

Whole body forward dynamics simulation models are typically either torque-driven or muscle-driven. Muscle-driven models enable the study of relatively complex systems (Neptune & Hull, 1998; Anderson & Pandy, 1999) and are necessary when the role of individual muscles are investigated (Pandy, Zajac, Sim, & Levine, 1990; van Soest, Schwab, Bobbert, & van Ingen Schenau, 1993; Jacobs, Bobbert, & van Ingen Schenau, 1996; Bohm, Cole, Bruggemann, & Ruder, 2006). Muscle-driven models typically require that many of the individual muscle strength parameters are selected from multiple literature sources and therefore may not in these cases be representative of either an individual or a generic human. In some cases, authors have looked to avoid introducing multiple sources of error into complex individual muscle driven models by attempting to establish a cohesive data set from which parameters may be derived from a single-source. In the case of the study by Arnold, Ward, Lieber, and Delp (2010) a data set was utilised which enabled multiple individual parameters to be collected from a single source where multiple muscle-tendon parameters were identified for the same cadaver. Their simulation model was intended to study the role and function of human muscle and tendon, which it was well-placed to achieve. However the same method is not appropriate for identifying a global optimum solution for movement

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since there is no kinematic and kinetic data against which to evaluate the model.

Some authors using an individual muscle modelling approach have sought to minimise simulation model errors by reducing the complexity of the design by grouping muscles together which have similar operational roles, for example by considering the three hamstrings as a single hamstring muscle (van Soest et al., 1993). This reduced parameter set might then be be scaled to represent an individual or small subset of participants (Domire & Challis, 2007), enabling a simulation model to investigate muscle and tendon roles whilst also retaining the possibility to evaluate the movement solution against participant data. If the predominant need for a simulation model is to investigate aspects of technique, with less dependence on understanding this technique at the level of individual muscle contributions, then a lumped muscle-modelling approach will provide the necessary solution and further reduce the complexity of the model and potential sources of error. Torque-driven models have the advantage that it is possible to determine subject-specific strength parameters from measurements made using an isovelocity dynamometer and to then evaluate the whole-body simulation model against a recorded performance (King, Wilson, & Yeadon, 2006; Allen, King, & Yeadon, 2013). In order that the model adequately produces a realistic performance, it is the generation of realistic net joint torques that is required, rather than any in depth understanding of the individual role of muscles. As a consequence fewer muscle and tendon parameters are required.

Historically torque-driven models have produced good representations for a number of activities (e.g. tumbling takeoff, King & Yeadon, 2004; high jumping takeoff, Wilson, Yeadon, & King, 2007; triple jumping takeoffs, Allen, King, & Yeadon, 2010), with the torque at a joint calculated only from single joint kinematics (monarticular representations). In these models the effect of biarticularity and changes in length at a secondary joint are ignored. Thus it is not clear what advantage incorporating biarticular representations may have and in which circumstances they may be beneficial. More recently, subject-specific combined monoarticular and biarticular torque generator parameters at the ankle (Lewis, King, Yeadon, & Conceição, 2012) and knee joint (King, Lewis, & Yeadon, 2012) have been derived from isovelocity torque measurements to address the issue of biarticular effects. For ankle plantar flexor joint torque it was demonstrated that if the knee was flexed by more than 40° a two-joint representation was appropriate (Lewis et al., 2012). Similarly, at the knee the total flexor and total extensor joint torque representations were improved when monoarticular and biarticular components were used as opposed to just using the knee joint kinematics alone (King et al., 2012). The additional complexity in this two-joint lumped approach, enables groups of muscles operating with similar function, to be simplified into two groups, those which are affected by one joint and those which are affected by two joints, but still without the detail being at the individual muscle level.

At the hip, previous subject-specific torque-driven forward-dynamics simulation models have calculated maximum voluntary hip flexor and extensor torques using the kinematics of the hip alone (e.g. tumbling; King & Yeadon, 2004, diving; King, Kong, & Yeadon, 2005, jumping for height; King et al., 2006). The monoarticular muscle contributions to total hip flexor torque and total hip extensor torque may be in the region of 62% and 46% respectively based upon physiological cross-sectional area, pennation angle and moment arms in the literature assuming equal activation of all muscles (Appendix A). Biarticular hip-knee muscles make a substantial contribution to torque at the hip joint, although it is not clear if including the biarticular contribution within a monoarticular representation at the hip is appropriate for all movements. The maximum knee flexor and knee extensor torques exerted by the two-joint knee-hip torque generators of King et al. (2012) can be used to calculate the torques which would be exerted by the same biarticular components at their secondary joint (in this case the hip) by using the ratio of moment arms previously established through optimisation (Appendix B). Fig. 1 shows the hip flexor and hip extensor torques generated by the biarticular knee extensor and biarticular knee flexor torque generators respectively. Here the activation is assumed to be maximal and the knee angle fixed at one of three joint angles throughout the functional joint range. Biarticular hip flexion and hip extension representations would contribute the largest hip torques when the biarticular muscles were at their longest lengths and contracting eccentrically (hip flexion: knee flexed, hip extended Fig. 1a, b, c; hip extension: knee extended, hip flexed, Fig. 1d, e, f). The biarticular hip torque would vary substantially as a function of knee angle; for biarticular hip flexion the eccentric torque would range from 14 Nm when the knee was flexed through to 0 Nm if the knee was within 42° of full extension (Fig. 1a, b, c). For biarticular hip extension the maximum eccentric torque would equate to 95 Nm, with biarticular torques unable to be exerted for posterior knee joint angles of 44° to full flexion (Fig. 1d, e, f). It is clear that were a biarticular knee-hip torque generator to be making a contribution to the net hip flexor or hip extensor torque, then the capacity of the hip joint to exert torque would differ from a single monoarticular representation of hip torques measured for a fixed knee joint angle if there were multi-joint kinematics.

Therefore the aim of this study was to establish the contribution of monoarticular hip flexors and hip extensors to total hip flexor and total hip extensor joint torques for an individual using previously derived parameters for biarticular knee-hip torque generators and to determine the magnitude of the difference between a single-joint and two-joint representation of hip torques.

2. Methods

2.1. Participants

Maximum voluntary isometric and isovelocity hip flexion and hip extension joint torques were measured for a single participant (28 year old male volleyball player, height 1.74 m, mass 79.2 kg) with experience of strength measurement on isovelocity dynamometers. The participant gave informed consent for the procedures in accordance with a protocol approved by the Loughborough University ethics committee.