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Contributions of muscle forces and toe-off kinematics to peak knee flexion during the swing phase of normal gait: an induced position analysis

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Abstract

A three-dimensional dynamic simulation of walking was used together with induced position analysis to determine how kinematic conditions at toe-off and muscle forces following toe-off affect peak knee flexion during the swing phase of normal gait. The flexion velocity of the swing-limb knee at toe-off contributed 30° to the peak knee flexion angle; this was larger than any contribution from an individual muscle or joint moment. Swing-limb muscles individually made large contributions to knee angle (i.e., as large as 22°), but their actions tended to balance one another, so that the combined contribution from all swing-limb muscles was small (i.e., less than 3° of flexion). The uniarticular muscles of the swing limb made contributions to knee flexion that were an order of magnitude larger than the biarticular muscles of the swing limb. The results of the induced position analysis make clear the importance of knee flexion velocity at toe-off relative to the effects of muscle forces exerted after toe-off in generating peak knee flexion angle. In addition to improving our understanding of normal gait, this study provides a basis for analyzing stiff-knee gait, a movement abnormality in which knee flexion in swing is diminished.

Keywords: Swing; Stiff-knee gait; Dynamic simulation; Induced accelerations; Induced positions; Cerebral palsy

1. Introduction

During normal gait, advancement of the swing limb is enabled by flexion of the swing knee. In stiff-knee gait, a movement abnormality observed in persons with cerebral palsy and individuals after stroke, knee flexion during swing is diminished. This inhibits toe-clearance, resulting in tripping or requiring energy-inefficient compensatory movements (Sutherland and Davids, 1993). The diminished knee flexion associated with stiff-knee gait is frequently attributed to abnormal activity of the rectus femoris (Perry, 1987; Sutherland et al., 1990). Accordingly, treatments, such as a rectus femoris transfer surgery (Gage et al., 1987) or injections of neuromuscular blocking agents (Sung and Bang, 2000), are performed to alter the function of this muscle. Unfortunately, not all patients benefit from these treatments. We believe that outcomes may be variable, in part, because factors other than abnormal rectus femoris excitation limit knee flexion in some cases. Understanding the factors that produce knee flexion during the swing phase of normal gait is needed to provide a basis for investigation of the causes of limited knee flexion in stiff-knee gait.

Electromyographic data shows that muscles are active during the swing phase of gait, even though this activity is low relative to stance phase (Winter, 1991; Perry, 1992). However, activation patterns alone do not elucidate how muscles contribute to knee motion during swing due to the complex dynamics of the lower limbs (Zajac and Gordon, 1989). Studies of stiff-knee gait have characterized the roles that swing-limb joint moments and muscles play in generating swing-phase knee flexion. Riley and Kerrigan (1998) used dynamic simulation of stiff-knee kinematics to show that an increase in the hip flexion moment during swing can

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increase knee flexion. A muscle-driven simulation of normal swing phase (Piazza and Delp, 1996) showed that either an increase in knee extension moment or a decrease in hip flexion moment decreases knee flexion during swing. By calculating the angular accelerations of the knee induced by individual muscles, they found that the rectus femoris acts to accelerate the knee into extension during swing phase, while the biceps femoris short head, hip flexors, and ankle dorsiflexors accelerate the knee into flexion.

Joint angular velocities at toe-off also contribute to swing-phase knee flexion. Dynamic simulations of swing phase performed in the absence of muscular joint torques have approximated normal knee kinematics when the initial joint angular positions and velocities were chosen judiciously (Mochon and McMahon, 1980; Mena et al., 1981). Piazza and Delp (1996) found that the amount of knee flexion achieved during swing phase could be decreased by either increasing hip flexion velocity at toe-off or decreasing knee flexion velocity at toe-off. Goldberg et al. (2003) found that many stiff-knee subjects with cerebral palsy exhibit abnormally low angular knee flexion velocities at toe-off, and that a simulated increase in this velocity results in a normal or above normal range of knee flexion in swing.

A balance between the factors that promote and inhibit knee flexion is required to achieve adequate knee flexion during swing; stiff-knee gait results when this balance is not achieved. However, the factors that contribute to this balance have not been clearly identified or quantified. The purpose of this study was to determine the contributions of individual muscles, joint moments, gravity, Coriolis and centripetal forces, and knee flexion velocity at toe-off to peak knee flexion during the swing phase of normal gait. In particular, we assessed the importance of knee flexion velocity at toe-off relative to the importance of muscle forces exerted after toe-off. These results add to our understanding of normal gait and point to factors that could enable more effective treatment of stiff-knee gait.

2. Methods

To assess the contributions of individual muscles and other factors to peak knee flexion in swing, we calculated induced positions. As the basis for our analysis, we used a dynamic simulation of normal gait (Anderson and Pandy, 2001) in which the body was modeled as a 10 segment, 23 degree-of-freedom linkage controlled by 54 musculotendon actuators. The first 6 degrees of freedom were used to define the position and orientation of the pelvis relative to the ground. The remaining nine segments branched out from the pelvis. The head, arms, and torso were represented as a single rigid body that articulated with the pelvis via a ball-and-socket joint located at approximately the third lumbar vertebra. Each hip was modeled as a ball-and-socket joint, each knee as a hinge joint, each ankle-subtalar joint as a universal joint, and each metatarsal joint as a hinge joint. Interactions of the feet with the ground were modeled using five viscoelastic elements distributed under the sole of each foot. Each actuator was modeled as a 3-element, Hill-type muscle in series with tendon (Zajac, 1989). The muscle–tendon parameters and origin and insertion sites were based on data reported by Delp et al. (1990). The effects of ligaments, modeled as angle-dependent torques, were included to prevent anatomically infeasible joint angles. See Anderson and Pandy (1999) for details concerning the model. The joint angular displacements, ground reaction forces, and muscle excitation patterns predicted using this model were similar to those measured from five healthy subjects (Anderson and Pandy, 2001).

The accelerations of the joints induced by the forces acting on the body were computed using the equations of motion for the model:

$$\vec{\ddot{q}} = \overleftrightarrow{I}^{-1}\left\{\vec{F}_G + \vec{F}_C + \vec{F}_L + \sum_{m=1}^{54}\vec{F}_m + \vec{F}_S\right\}, \tag{1}$$

where $\vec{\ddot{q}}$ is the vector of joint accelerations, $\overleftrightarrow{I}^{-1}$ is the inverse of the system mass matrix, and $\vec{F}_G$, $\vec{F}_C$, $\vec{F}_L$, $\vec{F}_m$ and $\vec{F}_S$ are vectors of generalized forces due to gravity (G), Coriolis and centripetal forces (C), ligaments (L), a muscle force (m), and ground reaction forces (S) that were modeled by the foot springs (Kane and Levinson, 1985; Zajac and Gordon, 1989). The vector $\vec{F}_m$ is obtained by multiplying the moment arm vector $\vec{R}_m$ for muscle m (i.e., the m^{th} column of the moment arm matrix $\overleftrightarrow{R}$) by the muscle force f_m in muscle m:

$$\vec{F}_m = f_m\,\vec{R}_m. \tag{2}$$

The ground reaction force was treated as a passive response to the other forces acting on the body and was decomposed by assuming rigid contact with the ground:

$$\vec{F}_S = \vec{F}_S^G + \vec{F}_S^C + \vec{F}_S^L + \sum_{m=1}^{54}\vec{F}_S^m + \vec{F}_S^I, \tag{3}$$

where $\vec{F}_S^G$, $\vec{F}_S^C$, $\vec{F}_S^L$, $\vec{F}_S^m$, and $\vec{F}_S^I$ represent the contributions of gravity, Coriolis and centripetal effects, ligaments, muscle m, and inertial forces, respectively, to the total ground reaction force $\vec{F}_S$ (Anderson and Pandy, 2003). The inertial force $\vec{F}_S^I$ is a fictitious force that is necessary to account for any non-rigid contact of the feet with the ground. It is large only when the feet come into or out of contact with the ground. Details concerning the decomposition methodology and the inertial force are presented in Anderson and Pandy (2003). The accelerations ($\vec{\ddot{q}}$) induced by a generalized force were then computed by multiplying the sum of the generalized force ($\vec{F}_i$) and its associated contribution to the ground reaction force ($\vec{F}^i_S$) by the inverse of the system mass matrix ($\overleftrightarrow{I}^{-1}$):

$$\ddot{\vec{q}}_i = \overleftrightarrow{I}^{-1}\{\vec{F}_i + \vec{F}^i_S\}, \quad i = G, C, L, m(1\ldots54), I. \tag{4}$$

When the ground reaction force is decomposed properly, the observed accelerations ($\ddot{\vec{q}}$) can be reconstructed by summing the individual induced accelerations:

$$\ddot{\vec{q}} = \sum_i \ddot{\vec{q}}_i, \quad i = G, C, L, m(1\ldots54), I. \tag{5}$$

Eq. (5) can then be integrated over a time interval (t_0 to t) to reconstruct the simulated joint velocities:

$$\dot{\vec{q}} = \dot{\vec{q}}_0 \sum_i \left\{ \int_{t_0}^{t} \ddot{\vec{q}}_i \, dt \right\}, \quad i = G, C, L, m(1\ldots54), I, \tag{6}$$

where $\dot{\vec{q}}_0$ is a vector of the initial joint velocities at time t_0. Eq. (6) can be integrated to reconstruct the simulated joint positions:

$$\vec{q} = \vec{q}_0 + \int_{t_0}^{t} \dot{\vec{q}}_0 \, dt + \sum_i \left\{ \int_{t_0}^{t} \int_{t_0}^{t} \ddot{\vec{q}}_i \, dt \, dt \right\},$$
$$i = G, C, L, m(1\ldots54), I, \tag{7}$$

where $\vec{q}_0$ is a vector of the initial joint positions at time t_0. The changes in joint positions induced by a force component i over the time interval t_0 to t are then given by

$$\vec{q}_i \equiv \int_{t_0}^{t} \int_{t_0}^{t} \ddot{\vec{q}}_i \, dt \, dt. \tag{8}$$

Using this definition of induced positions (Eq. (8)) and realizing that $\dot{\vec{q}}_0$ is a constant, Eq. (7) can be simplified to yield the following expression:

$$\vec{q}(t) = \underbrace{\vec{q}_0}_{\substack{\text{initial}\\ \text{positions}}} + \underbrace{\dot{\vec{q}}_0(t - t_0)}_{\substack{\text{changes in positions}\\ \text{due to}\\ \text{initial velocities}}} + \underbrace{\sum_i \vec{q}_i}_{\substack{\text{induced positions}\\ \text{due to}\\ \text{applied forces}}},$$
$$i = G, C, L, m(1\ldots54), I. \tag{9}$$

Thus, the changes in the simulated joint positions $\vec{q}_0(t)$ that occur over a time interval t_0 to t can be explained by contributions from the initial joint velocities and from the sum of the induced positions due to the forces acting on the body.

To quantify the contribution of forces and initial conditions to peak knee flexion during swing, the terms in Eq. (9) were evaluated for the knee joint angle of the swing limb from the time of toe-off ($t_0 = 0.0\,\mathrm{s}$) to the time of peak knee flexion ($t = 0.08\,\mathrm{s}$):

$$q_k(t) = \underbrace{q_{k_0}}_{\substack{\text{initial knee}\\ \text{angle}}} + \underbrace{\dot{q}_{k_0} \cdot (t - t_0)}_{\substack{\text{knee angle}\\ \text{due to initial}\\ \text{knee velocity}}} + \underbrace{\sum_i \int_{t_0}^{t} \int_{t_0}^{t} \ddot{q}_{k_i} \, dt \, dt}_{\substack{\text{knee angle}\\ \text{induced by}\\ \text{component } i}},$$
$$i = G, C, L, m(1\ldots54). \tag{10}$$

During this time interval, the stance foot was in near rigid contact with the ground; therefore, $\vec{F}^I_S$ was negligibly small and was omitted from the analysis. Thus, the factors that were considered to contribute to peak knee flexion during swing were the initial knee angle at toe-off (q_{k_0}), the initial knee flexion velocity at toe-off ($\dot{q}_{k_0}$), gravitational forces, Coriolis and centripetal forces, the ligament torques, and the muscle forces. The contributions to peak knee flexion made by the net joint moments were also calculated by replacing the ligament (L) and muscle (m) terms in Eq. (4) with the appropriate joint moments.

3. Results

The initial angular velocity of the swing knee made the largest contribution to peak knee flexion. The initial knee flexion velocity ($375^\circ\,\mathrm{s}^{-1}$) contributed 30° of knee flexion to this change in angle (Fig. 1, dashed black line). The forces applied by all actuators together (actuators include muscles and ligaments) acted to extend the knee by 12° (Fig. 1, thick black line). Gravitational, Coriolis, and centripetal forces each had little influence on peak knee flexion angle; when combined these forces acted to

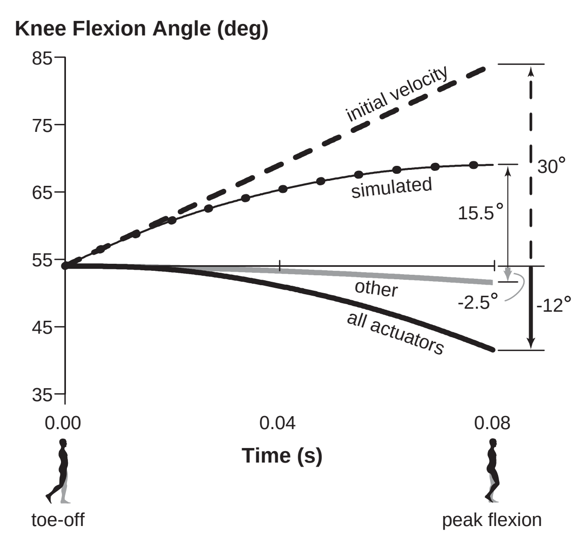

Fig. 1. Contributions of forces and toe-off kinematics to the observed motion of the swing-limb knee from the time of toe-off to the time of peak knee flexion. Flexion is positive. During the simulation, the knee flexed from an initial angle of 54° to a peak knee flexion angle of 69.5° for a total change of 15.5° (thin black line). The initial knee flexion velocity (dashed black line) contributed 30° of flexion to the observed motion. All actuators (muscles and ligaments; thick black line) contributed 12° of extension. All other forces (i.e., gravitational, Coriolis, and centripetal forces; gray line) made small contributions which, when combined, amounted to only 2.5°. When the contributions from initial knee flexion velocity, actuators, and all other forces are summed (black circles), the observed motion of the knee is reconstructed.

extend the knee by less than 3° (Fig. 1, gray line). The knee joint angle induced by the combination of all factors overlays the actual trajectory of the knee joint angle (Fig. 1, compare thin black line to black circles). This condition is necessary for the induced position analysis to be valid (see Eq. (9)).

Both the swing-limb muscles and the back muscles acted to flex the knee by about 3° (Fig. 2, thin black and thin gray lines). The stance-limb muscles acted to extend the knee by 18° (Fig. 2, dashed black line). The muscles that contributed most to this net extension were the stance-limb gluteus maximus and the stance-limb gluteus medius/minimus.

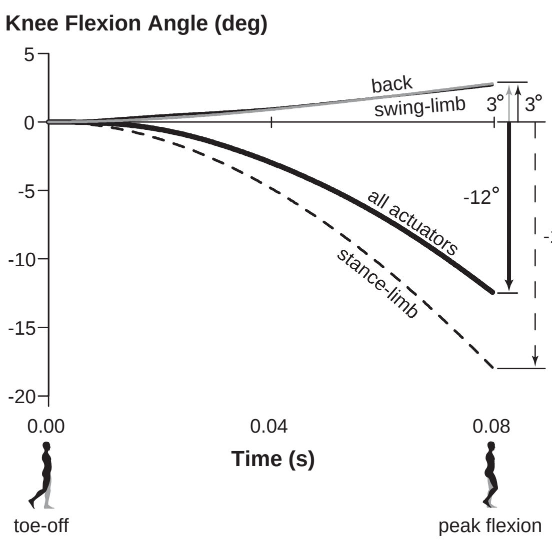

Fig 2. Contributions of the stance-limb muscles, swing-limb muscles, back muscles, and the combination of all actuators to the observed motion of the swing-limb knee from the time of toe-off to the time of peak knee flexion. Flexion is positive. All actuators combined (muscles and ligaments; thick black line) contributed 12° of extension to peak knee flexion angle. The swing-limb muscles (thin black line) and back muscles (thin gray line) each made only small contributions (i.e., only about 3° of flexion). The stance-limb muscles (dashed black line), because they acted to support the swing-limb hip, made a larger contribution of 18° of extension.

The individual muscles that made the largest contributions to peak knee flexion were the swing-limb vasti (due predominantly to passive forces), ankle dorsiflexors, iliopsoas, and uniarticular ankle plantarflexors, and the stance-limb gluteus medius/minimus and gluteus maximus (Fig. 3a). The biarticular muscles (rectus femoris, gastrocnemius, and hamstrings) each contributed less than 3° to knee angle (Fig. 3b). The swing-limb net joint moments that most influenced peak knee flexion were, in descending order, knee extension, hip flexion, and ankle dorsiflexion (Fig. 4).

4. Discussion

We used induced positions to quantify how knee flexion velocity at toe-off and the forces due to muscles, gravity, and Coriolis and centripetal effects contribute to peak knee flexion angle during the swing phase of normal gait. Induced positions are an extension of the induced accelerations concept (Hollerbach and Flash, 1982; Zajac and Gordon, 1989). Induced accelerations have been used to understand the muscle coordination of a variety of movements, including jumping, pedaling, and walking (e.g., Pandy et al., 1990; Fregly and Zajac, 1996; Neptune et al., 2001; Zajac, 2002). However, induced accelerations have not been used to quantify the contributions of forces to joint positions. The difficulty in interpreting how induced accelerations contribute to position arises because accelerations are instantaneous

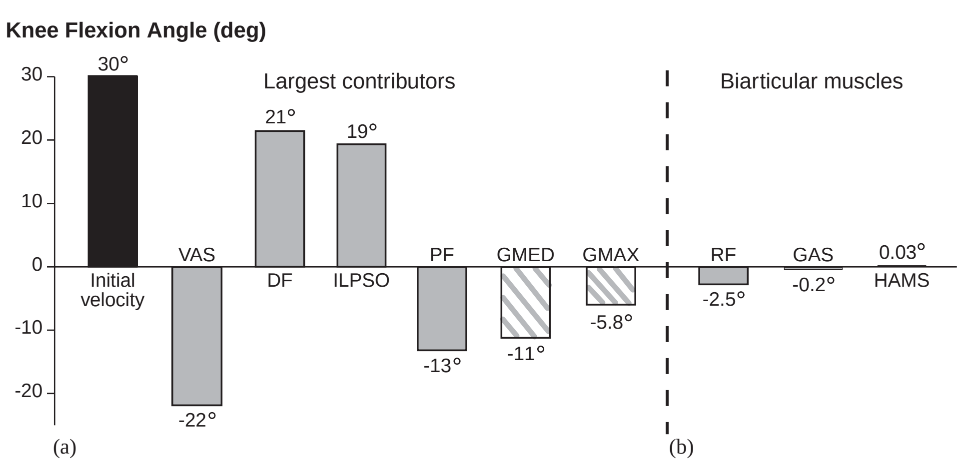

Fig 3. Contributions of toe-off knee flexion velocity and individual muscles to peak knee flexion angle. Flexion is positive. Swing-limb muscles are shown as solid and stance-limb muscles are show as hashed. (a) Largest contributors. The largest contributions to peak knee flexion angle from individual muscles were made by the swing-limb vasti (VAS), ankle dorsiflexors (DF), iliopsoas (ILPSO), and uniarticular ankle plantarflexors (PF), and the stance-limb gluteus medius/minimus (GMED) and gluteus maximus (GMAX). Each of these contributions was smaller than the contribution that resulted from the flexion velocity of the knee at toe-off (30°; Initial velocity). (b) Biarticular muscles. The contributions to peak knee flexion angle from the swing-limb rectus femoris (RF), gastrocnemius (GAS), and hamstrings (HAMS) were small.

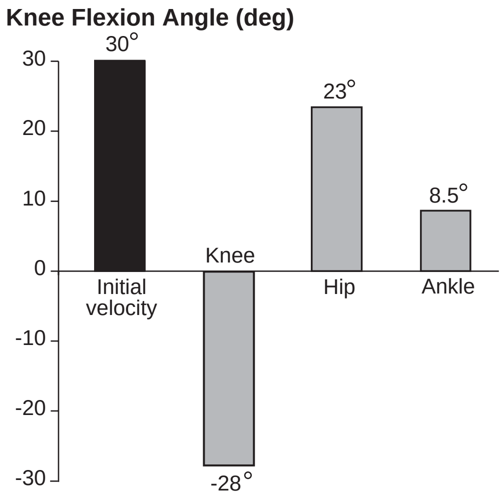

Fig. 4. Contributions of toe-off knee flexion velocity and net swing-limb joint moments to peak knee flexion angle. Flexion is positive. The ankle moment contributed less than either the hip or knee moments. The knee extension moment contributed almost as much as the initial knee flexion velocity, although in the opposite direction.

quantities. They quantify the magnitude and direction in which a force accelerates a joint at a particular time, but their contribution to movement depends also on the time interval over which they are generated. More explicitly, position is related to acceleration by a double integral over time, and this is the relation we use to define an induced position (Eq. (8)). As a consequence of this definition, given an interval of movement, a force that is applied early in the interval will likely be more influential than a force that is applied late in the interval. Induced positions quantify such effects by accounting for the time histories of applied forces and the accelerations they produce.

Consistent with the findings of others (Mochon and McMahon, 1980; Mena et al., 1981; Kerrigan and Glenn, 1994; Piazza and Delp, 1996; Goldberg et al., 2003), our induced position analysis showed that knee flexion velocity at toe-off made the largest contribution to peak knee flexion during swing. The initial knee flexion velocity in the simulation ($375^\circ\,s^{-1}$) was somewhat larger than that used by Piazza and Delp (1996) ($322^\circ\,s^{-1}$). However, the subjects in Piazza and Delp (1996) walked at a somewhat slower average speed ($1.17\,m\,s^{-1}$) than the model ($1.38\,m\,s^{-1}$). Based on the formulation of induced positions (Eq. (9)), if the time to peak knee flexion angle remained the same but the initial knee flexion velocity were $322^\circ\,s^{-1}$, our estimate for the contribution of initial knee flexion velocity would decrease from 30° to 25°. This contribution would still be greater than any other individual muscle (Fig. 3a) and second in magnitude only to the contribution from the net swing-limb knee extension moment (Fig. 4).

The substantial contribution of the stance-limb muscles to peak knee angle occurred primarily because the stance-limb muscles support the swing-limb hip. Of the 18° of knee extension contributed by the stance-limb muscles (Fig. 2), 11° were contributed by the stance-limb gluteus medius/minimus (Fig. 3a), which are important for controlling pelvic list and providing vertical support (Inman, 1947; Anderson and Pandy, 2003). Gravity did not contribute substantially to peak knee angle because passive support of the bones and joints against gravity did little to support the swing limb. While the stance leg received substantial support against gravity (i.e., about 30% of body weight; Anderson and Pandy, 2003), the swing limb, including the swing-limb side of the pelvis, was in near free-fall. Because all the segments of the swing limb accelerated downward nearly uniformly, gravity did little to induce accelerations in the swing-limb knee. If the simulation of swing had been conducted using a model of only the swing limb in which the trajectory of the hip was prescribed, gravity, rather than the stance-limb muscles, would have contributed substantially to knee extension.

In the simulation there was a balance between the opposing actions of the swing limb muscles. For example, the vasti acted to extend the knee by 22°, but the iliopsoas acted to flex the knee by 19° (Fig. 3a). Consequently, in net, the swing-limb muscles acted to flex the knee by only 3° (Fig. 2). Piazza and Delp (1996), who considered only swing-limb muscles, reported that the total muscle-induced acceleration of the knee was in the direction of extension. This inconsistency may be due to the fact that the motion of the swing hip was prescribed in the simulations conducted by Piazza and Delp (1996). It may also have arisen from particular features of the dynamic optimization solution obtained by Anderson and Pandy (2001). It is possible that the swing-limb muscles act to extend the knee in some gait patterns and flex it in others. The more robust result, we believe, is that the net contribution of swing-limb muscles to peak knee flexion in normal gait is small.

The most influential muscles were the swing-limb vasti, iliopsoas, dorsiflexors, and plantarflexors (Fig. 3a). The large contributions of vasti and iliopsoas were expected in light of previous findings that hip and knee moments strongly influence peak knee flexion (Piazza and Delp, 1996; Riley and Kerrigan, 1998). However, our findings that the dorsiflexors and uni-articular plantarflexors made large contributions to peak knee flexion is somewhat surprising.

Activation of the dorsiflexors is needed to keep the ankle dorsiflexed during swing (Mena et al., 1981; Perry, 1992). Our results indicate that the dorsiflexors also act to flex the knee during early swing (Fig. 3a). During this period, the action of the dorsiflexors was directly opposed by the passive forces generated by the plantar-flexors. In net, the swing-limb ankle moment changed peak knee flexion angle by only 8.5° (Fig. 4). This level of influence on knee angle is compatible with the finding of Piazza and Delp (1996) that ankle moments were substantially less influential than either hip or knee moments.

The contributions to peak knee flexion angle from rectus femoris, hamstrings, and gastrocnemius were all less than 2.5°, nearly 10 times less than the contributions from vasti or iliopsoas (Fig. 3). The explanation for the relatively small contributions of these biarticular muscles is two-fold. First, they did not develop large forces in the simulation during early swing. Second, the contribution to knee flexion from the moment exerted at one joint opposed the contribution from the moment exerted at the second joint. For example, the hip flexion moment generated by rectus femoris acted to flex the knee, like iliopsoas; however, the knee extension moment generated by rectus femoris acted to extend the knee, like vasti.

The diminished knee flexion associated with stiff-knee gait is commonly attributed to over-activity of the rectus femoris during swing, and treatment is often aimed at altering the function of this muscle (Perry, 1987). To understand the geometric factors that contribute to the efficacy of rectus femoris tendon transfers, we normalized the induced position given in Eq. (10) by the time history of muscle force:

$$\tilde{q}_{k_m}(t) \equiv \int_{t_0}^{t} \int_{t_0}^{t} \frac{\ddot{q}_{k_m}}{f_m} \, dt \, dt, \tag{11}$$

for sartorius, iliopsoas, biceps femoris short head, rectus femoris, and hamstrings. The contribution per unit muscle force to knee angle by rectus femoris was less than that by most other muscles (Fig. 5). From this, we conclude that an intervention that reduces the force output of rectus femoris would have less impact than one that transforms rectus femoris into a uniarticular hip flexor, like iliopsoas, or even better, into a biarticular hip flexor/knee flexor, like sartorius. While rectus femoris transfer surgery is often intended to convert the rectus femoris into a biarticular hip flexor/knee flexor like sartorius, recent studies have shown that this procedure does not generally achieve this result (Riewald and Delp, 1997; Asakawa et al., 2002). The procedure may still be effective in enabling knee flexion by reducing or eliminating the muscle's capacity to generate a knee extension moment while preserving its capacity to generate a hip flexion moment.

Induced positions provide an intuitive measure for systematically characterizing the contributions of forces to movement. They provide a measure in degrees of how individual muscles contribute to an observed joint displacement and allow the effects of forces and velocities to be compared directly. However, there are a number of limitations. First, while induced positions provide a precise accounting for how forces contribute to an observed motion, they provide only a limited basis for predicting the movement consequences of altered

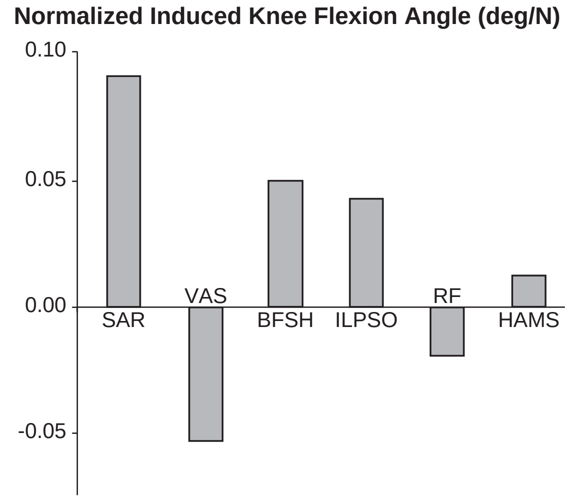

Fig. 5. The normalized induced positions of swing-limb muscles that cross only the hip and/or knee: sartorius (SAR), vasti (VAS), biceps femoris short head (BFSH), iliopsoas (ILPSO), rectus femoris (RF), and hamstrings (HAMS). The normalized induced position due to a muscle is calculated by dividing the muscle-induced acceleration by the instantaneous force in the muscle and twice integrating (see Eq. (11)). Thus, the normalized induced position due to a muscle is primarily a function of muscle geometry and represents the geometric advantage of a muscle to contribute to peak knee flexion. Of the muscles considered, sartorius, a biarticular hip flexor/knee flexor, had the highest geometric advantage to contribute to peak knee flexion; rectus femoris and hamstrings had the lowest.

muscle activity. For example, it would be incorrect to conclude that doubling the force output of the dorsiflexors would double their contribution to peak knee flexion (i.e., 21° would not double to 42°, Fig. 3a). Second, induced positions depend on the degrees of freedom included in a model. This dependency arises not from the formulation of induced positions (Eq. (9)), but rather from how induced accelerations are computed. For example, if the trajectory of the swing hip is prescribed, constraint forces will eliminate any induced accelerations of the position of the swing hip and will also alter the induced accelerations of the rest of the joints in the limb. While such simplifying assumptions are often justifiable and desirable, care should be taken to understand their impact on results. Finally, induced positions quantify a change in position over a specified time interval. Thus, reconstruction of the observed kinematics depends on having a set of initial positions and velocities (Eq. (9)). If the contributions of the initial conditions to the observed kinematics are important, one is still left with the task of explaining how the initial conditions came about.

Calculating induced velocities and positions during double support is a natural next step toward understanding the origins of swing phase initial conditions. As in the current analysis, this requires a decomposition of the ground reaction force (see Eq. (3)). During swing phase, we have verified that our induced position results are insensitive to the method used to decompose the ground reaction force. During double support, however, we have found that results are sensitive to the decomposition methodology. Unfortunately, a definitive method for decomposing the ground reaction force has not yet been identified (Neptune et al., 2001; Anderson and Pandy, 2003). Our future work will focus on identifying a definitive decomposition methodology and on developing alternative methods for quantifying the factors during double support that give rise to the knee angle and knee angular velocity at toe-off.

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