

Predicting the Functional Roles of Knee Joint Muscles from Internal Joint Moments

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ABSTRACT

FLAXMAN, T. E., T. ALKJÆR, E. B. SIMONSEN, M. R. KROGSGAARD, and D. L. BENOIT. Predicting the Functional Roles of Knee Joint Muscles from Internal Joint Moments. *Med. Sci. Sports Exerc.*, Vol. 49, No. 3, pp. 527–537, 2017. **Introduction:** Knee muscles are commonly labeled as flexors or extensors and aptly stabilize the knee against sagittal plane loads. However, how these muscles stabilize the knee against adduction–abduction and rotational loads remains unclear. Our study sought 1) to classify muscle roles as they relate to joint stability by quantifying the relationship between individual muscle activation patterns and internal net joint moments in all three loading planes and 2) to determine whether these roles change with increasing force levels. **Methods:** A standing isometric force matching protocol required subjects to modulate ground reaction forces to elicit various combinations and magnitudes of sagittal, frontal, and transverse internal joint moments. Surface EMG measured activities of 10 lower limb muscles. Partial least squares regressions determined which internal moment(s) were significantly related to the activation of individual muscles. **Results:** Rectus femoris and tensor fasciae latae were classified as moment actuators for knee extension and hip flexion. Hamstrings were classified as moment actuators for hip extension and knee flexion. Gastrocnemius and hamstring muscles were classified as specific joint stabilizers for knee rotation. Vastii were classified as general joint stabilizers because activation was independent of moment generation. Muscle roles did not change with increasing effort levels. **Conclusions:** Our findings indicate muscle activation is not dependent on anatomical orientation but perhaps on its role in maintaining knee joint stability in the frontal and transverse loading planes. This is useful for delineating the roles of biarticular knee joint muscles and could have implications in robotics, musculoskeletal modeling, sports sciences, and rehabilitation. **Key Words:** KNEE JOINT STABILITY, MUSCLE ACTIVITY, LOWER LIMB KINETICS, HEALTHY CONTROLS, PARTIAL LEAST SQUARES REGRESSION

Because of their anatomical orientation, the major muscles crossing the knee are traditionally classified as knee joint flexors (hamstring and gastrocnemius) and extensors (quadriceps), and their corresponding activation is typically classified as either agonistic or antagonistic to sagittal plane motion. However, functional loads are not limited to a single axis. Theoretically, each muscle crossing the knee has a unique orientation so that when activated, it can generate a moment along a specific line of action and oppose the given external force causing that moment (4,11). However, because of their small cross-sectional area, muscles with large frontal plane moment arms, such as the gracilis, sartorius, and tensor fasciae latae (TFL), have low force generation capacity and minimal contribution to opposing

frontal plane external loads (9,11). It remains unclear how the muscles crossing the knee effectively stabilize the joint when frontal and rotational loads are applied.

In vivo evaluations of neuromuscular function with respect to knee joint stability typically involve a dynamic assessment (e.g., cutting maneuver, perturbation, etc.) such that results are functionally comparable with activities of sport and daily living. Yet the presence of biomechanical factors such as joint velocity, position, and direction of movement confounds the muscular contribution to force generation (41). To better elucidate this relationship, isometric exercises are commonly used. Previously, studies have limited such evaluations to a single axis (i.e., flexion–extension) and/or to a non–weight-bearing condition (3,11,38). Because noncontact knee joint injuries occur when the foot is in contact with the ground and the individual bears weight, a study that simulates physiological loads in a controlled yet more functionally relevant manner is critical to gain a better understanding of the neuromuscular contribution to force generation and joint stability.

We have developed a task that requires subjects to stand and modulate ground reaction forces (GRF) while maintaining a static position (16,17,37). Our results indicate that when standing and bearing body weight, a muscle's activation is not

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dependent on moment arm orientation (MAO). Rather, we propose that a muscle's activation is also related to its role in maintaining knee joint stability. We have classified the roles of knee muscles as 1) a general joint stabilizer: a muscle that demonstrates activation independent of loading directions; 2) a moment actuator: a muscle that demonstrates preferential activation in a loading direction corresponding to its MAO; and 3) a specific joint stabilizer: a muscle that demonstrates preferential activation in a loading direction opposite of its MAO (e.g., the biceps femoris activation pattern reported by Flaxman et al. [16,17] was specific about a medial loading direction, which is opposite of its posteriolateral MAO).

The purpose of this study was to further classify the roles of lower limb muscle as it relates to joint stability by quantifying the relationship between individual muscle activation patterns and internal net joint moments in all three loading planes (aim 1). This is especially beneficial when trying to determine the function of biarticular muscles spanning the hip and knee: which joint is the muscle more likely to act on? On the basis of our previous work (16,17) and that of van Ingen Schenau et al. (39), who suggested that monoarticular muscles provide a general gross force whereas biarticular muscles “fine tune” the distribution of the net joint moments across adjacent joints, it is hypothesized that 1) the internal net joint moments will be independent of changes in the activation of the vastii muscles and 2) the activation of the biarticular knee/hip muscles will be load dependent (significantly correlated by a general moment at each articulating joint), corresponding to their role as moment actuators.

Under isometric conditions, the relationship between muscle activation and force is frequently linear (6). Yet again, this relationship has been established from non-weight-bearing conditions and is limited to an analysis of a single axis (3,11,38). When comparing different populations, varied muscle activation characteristics are often accompanied with varied moment magnitudes (3,25). As such, we also sought to determine whether muscle activation patterns, and the roles of muscles as they relate to knee joint stability, will change with an increase in GRF level (aim 2). On the basis of works by Buchanan et al. (10) and Levin et al. (26), it is hypothesized that there will be a progressive increase in EMG amplitude with increasing force demand but a muscle's functional role will not change. With this, we can determine whether observed between-group differences indicate fundamental changes in neuromuscular control or simply a function of loading level.

METHODS

Subjects and equipment. Twenty-five healthy active adults (12 males, height = 182.9 ± 4.9 cm, mass = 81.2 ± 10.1 kg, age = 29.2 ± 6.7 yr, and 13 females, height = $169.7.9 \pm 5.0$ cm, mass = 63.1 ± 9.0 kg, age = 25.6 ± 7.3 yr) with no previous reports of significant lower limb injury were recruited from the University of Copenhagen and the surrounding community. Experiments were conducted at the

University of Copenhagen, and all subjects read and signed an informed consent form approved by the local ethics committee for the Capital Region of Denmark (De Videnskabetiske Komiteer for Region Hovedstaden, H-3-2013-126) and the University of Ottawa Research Ethics Board (H06-14-27).

Retroreflective markers were placed on various anatomical landmarks on the subject's body according to the HMBL Cluster Marker set ([30]; Fig. 1A). Trajectories were recorded using a 10-camera motion analysis system (6 MXF-40s and 4T40-series cameras; Oxford Metrics, Oxford, UK) sampling at 100 Hz with supporting Nexus software (version 1.8, Oxford Metrics). GRF recorded from a force platform (AMTI-OR6; AMTI, Watertown, MA) were also collected in Nexus at 1000 Hz and amplified with an internal gain of 1000. Bipolar surface electrodes collected EMG of rectus femoris (RF), vastus medialis, vastus lateralis, semitendinosus (ST), biceps femoris (BF), medial gastrocnemius (MG), lateral gastrocnemius (LG), TFL, gluteus medius (GM), and adductor (ADD) muscle group of the dominant leg (defined as leg used to kick a soccer ball as far as possible). EMG was sampled at 1000 Hz with a 20- to 500-Hz bandwidth and a 6-dB per octave filter slope recorded using a wireless EMG system (MQ air; Marq Medical, Farum, Denmark).

Maximum voluntary isometric contractions were performed before the experimental protocol. Hip flexion, hip extension, hip abduction, and hip adduction exercises were performed while standing in neutral position (no hip or knee flexion), and effort was exerted against a strap placed above the ankle. For plantarflexion, subjects stood and raised to their toes while resisting upward motion using wall mounted bars. Knee extension and flexion exercise was performed with manual resistance from the researcher while subjects sat with hip flexed to 90° and knee to 30° . Visual feedback and verbal encouragement were provided. EMG_{max} for each muscle was computed as a 50-ms mean about the maximum value in the conditioned EMG signal across all exercises.

Protocol. A force matching protocol assessed muscle activation patterns of subjects while they modulated GRF with their dominant limb (16,17,37). Subjects maintained a staggered standing position during testing such that their feet were spaced hip width apart and their test leg had approximate joint angles of 30° hip flexion, 30° knee flexion, and 10° ankle plantarflexion (Fig. 1A). A projected image of a cursor and a force target was placed in front of the subject, providing visual feedback of the direction and magnitude of the subject's force applied to the force platform from the test leg. The cursor moved with three degrees of freedom: 1) anterior/posterior loads (force along the $\pm y$ axis) moved the cursor upward/downward, 2) medial/lateral loads (force along the $\pm x$ axis) moved the cursor to the left/right, and 3) inferior/superior loads (force along the $\pm z$ axis) decreased/increased the size of the cursor.

The successful match of the cursor over the target for 0.5 s triggered simultaneous recording of 3D marker trajectories, GRF, and EMG. Twelve different target directions, evenly spaced by 30° about a circular trajectory (representative of various

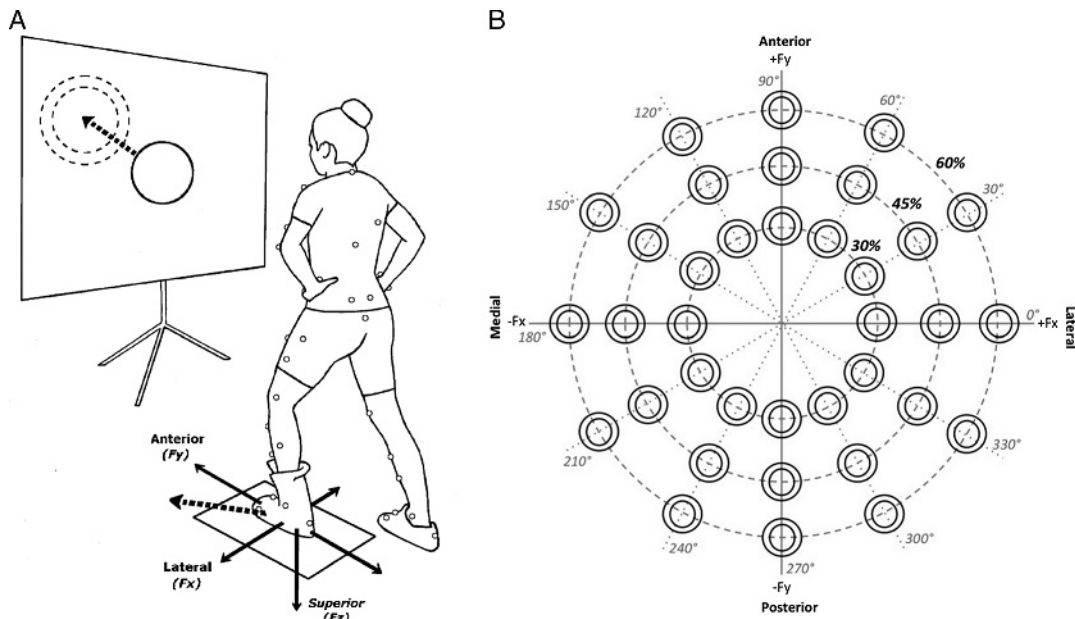


FIGURE 1—A, Subject stands with their dominant foot in a boot fixed to a force platform and the opposite foot located posteriorly and adjacently. A projector displays biofeedback of the applied forces as a cursor (solid circle). Subjects must position the cursor between the target's two rings (dashed circles) by modulating the force in the horizontal plane (F_x and F_y). Cursor diameter increases or decreases with more or less body weight loading (F_z), respectively. Example for a force trajectory required to match cursor to target is depicted as a dashed arrow. B, Force target locations representative of normalized horizontal force loading directions at three different force levels. Location along the radius represents normalized effort (30%, 45%, and 60%). Target directions are separated by 30° about a circular trajectory corresponding to the horizontal loading plane.

horizontal loading directions: 0° = lateral, 90° = anterior, 180° = medial, 270° = posterior), randomly appeared. Each direction had three different force magnitudes (30%, 45%, and 60% max effort). This set of 36 targets was matched twice for 72 targets total (Fig. 1B).

The effort required to successfully complete the force matching tasks was normalized to each participant's maximal standing force. The maximal standing force required participants to stand in the testing position, maintain equal body weight on each leg, and exert as much force on the force plate as possible with their test limb in the anterior, posterior, medial, and lateral anatomical loading directions corresponding to the respective $-F_y$, $+F_y$, $-F_x$, and $+F_x$ force plate channels. Although the subject maintained equal body weight on each leg, relaxed horizontal GRF were also recorded and subsequently subtracted from experimental GRF readings, acting as a “zeroing” method. The normalized force required to reach each target (F_{target}) (was computed based on Equation 1 presented in (25). This equation has been adapted to include the removal of the relaxed horizontal GRF:

$$F_{\text{target}} = \sqrt{[(\cos \Theta \times (F_{x_p} - F_{x_r}) \times \% \text{max} F)^2 + (\cos 90 - \Theta \times (F_{y_p} - F_{y_r}) \times \% \text{max} F)^2]}$$

where Θ is the angle between the target and the $+x$ axis, F_{x_p} and F_{x_r} are the respective peak and relaxed GRF produced along the $\pm x$ axis (medial/lateral), F_{y_p} and F_{y_r} are the respective peak and relaxed GRF produced along the $\pm y$ axis (anterior/posterior), and $\% \text{max} F$ is percent effort level.

Data processing. Raw marker trajectories and GRF were both filtered with a second-order 15 Hz dual-pass low-pass Butterworth filter and exported to OpenSim (v 3.2 [14]) to

compute local coordinate systems of the lower limb and subsequent joint angles and internal net joint moments using inverse kinematics and dynamics. Any future mention of the term “moments” refers to internal net joint moments, unless otherwise specified.

All EMG signals were high-pass filtered at 25 Hz with a second-order dual-pass Butterworth filter, full wave rectified, and low-pass filtered at 10 Hz with a second-order dual-pass Butterworth filter. EMG was normalized to maximal amplitude recorded from maximum voluntary isometric contractions ($\text{EMG}/\text{EMG}_{\text{max}}$). EMG, kinematic, and kinetic data were time averaged for the 0.5 s of successful force match, ensemble averaged across repetitions, and plotted in polar coordinates. To better represent the change in GRF effort on EMG and joint moments, relaxed EMG and joint moments recorded during the system's GRF “zeroing” step was also subtracted from corresponding trial data. In other words, the joint loads and EMG from the resting standing position were treated as a baseline and were not included in our graphs or analyses.

Group mean hip, knee, and ankle flexion angles were applied to an OpenSim musculoskeletal model (21) to extract the flexion–extension and adduction–abduction moment arms of muscles crossing the knee or hip during these positions. To quantify the orientation of a muscle's MAO with respect to a joint's axis of rotation, the \tan^{-1} of the Cartesian coordinates (adduction–abduction as x axis and flexion–extension as y axis) were computed for each muscle, 0° indicating pure abduction, 90° pure extension, 180° pure adduction, and 270° a pure flexion MAO.

Data analysis. The between-subject reliability of muscle activation and moment profiles was evaluated with intraclass correlation coefficients ($ICC_{(2,k)}$) (31) in the Statistical Package for the Social Sciences (version 18.0; IBM, Armonk, NY). To facilitate the between-subject comparison of the profiles in the ICC analysis, EMG was scaled to maximum value recorded in the given muscle of each subject.

Roles of individual muscles at each effort level were classified with symmetry analysis, mean direction of activation (Φ_M), and mean magnitude of activation (X_{EMG}) (16,17). A mean direction of moment generation (Φ_T) at 60% effort was also computed for each joint moment. *Moment actuators* were muscles whose muscle activation was asymmetrical about the polar plot origin, and Φ_M was not statistically different from the extracted MAO. *General joint stabilizers* were classified as muscles with symmetrical activation about the polar plot origin. *Specific joint stabilizers* were muscles with asymmetrical activation whose Φ_M was statistically different than its extracted MAO. Significant differences between a muscle's Φ_M and their extracted MAO were tested with a one sample test for the mean angle at the $\alpha < 0.05$ level (CircStat Toolbox for Matlab [5]).

To determine whether muscle activation patterns, and the roles of muscles as they relate to knee joint stability, will change with an increase in GRF level (aim 2), a one-way repeated-measures ANOVA was used for each muscle to test if X_{EMG} significantly differed across the factor of effort (three levels: 30%, 45%, and 60%) (Statistical Package for the Social Sciences, version 18.0, IBM) (17). The circular analog of an ANOVA, the Watson–Williams test (CircStat Toolbox for Matlab [5]), tested if Φ_M for each muscle significantly differed across the factor of effort (17).

The relationship between individual muscle activations and internal net joint moments across all three loading planes (aim 1) was evaluated with a partial least square regression (PLSR) analysis (XLSTAT, New York, NY) with cross validation (leave-one [subject]-out method) (20,43). This method is appropriate when multicollinearity is present among the predicting variables. The predictor variables (i.e., internal joint moments) are combined into principle components (PC) and then regressed onto the dependent variable (muscle activation) using ordinary least squares. Results are transformed back onto the normal X scale to obtain estimates of each predictor's standardized coefficients (β). The number of PC used was determined by accounting for 95% of total variance observed, or if adding another PC contributed to less than 5% VAF. The accuracy of model predictions was assessed with the variance explained (R^2) statistic. We classified a very weak prediction accuracy as $R^2 < 0.1$, weak as $0.1 < R^2 < 0.3$, moderate as $0.3 < R^2 < 0.5$, and strong as $R^2 > 0.5$. A predictor's standardized coefficient described the relative increase in muscle activation (EMG/EMG_{max}) with an increase in a given moment of $1 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ when all other moments are held constant. For example, a β of +0.2 for knee flexion indicates that a muscle's activation is predicted to increase by 20% of EMG_{max} when there is an increase of

$1 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ of knee flexion. Only predictors with significant positive coefficients were considered meaningful. Significance of β values was determined with 95% confidence intervals.

RESULTS

On the basis of the group mean sagittal plane hip, knee, and ankle joint angles, the MAO of muscles crossing the hip and knee joints are presented in Figures 2A and 3. After removing the relaxed stance forces, group mean \pm SD maximum normalized force levels in the anterior, posterior, medial, and lateral loading directions were 1.67 ± 0.51 , 2.46 ± 0.69 , 1.35 ± 0.39 , and $1.37 \pm 0.41 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$, respectively. Group mean \pm SD normalized force required to move the cursor to 60% effort at 90° (anterior) was $1.05 \pm 0.31 \text{ N}\cdot\text{kg}^{-1}$, 270° (posterior) $1.50 \pm 0.43 \text{ N}\cdot\text{kg}^{-1}$, 0° (lateral) $0.82 \pm 0.29 \text{ N}\cdot\text{kg}^{-1}$, and 180° (medial) $0.81 \pm 0.24 \text{ N}\cdot\text{kg}^{-1}$ (Fig. 2B). Respective hip flexion, knee flexion, and ankle plantarflexion mean \pm SD joint angles were $28.4^\circ \pm 6.7^\circ$, $23.2^\circ \pm 5.8^\circ$, $5.9^\circ \pm 4.8^\circ$ and normally distributed across subjects (Shapiro–Wilk $P > 0.05$).

Muscle activation. For all muscles, a significant increase in X_{EMG} was observed with an increase in effort level (Fig. 2C). EMG polar plots (Fig. 3) depict individual muscle activation patterns at all three effort levels, group mean Φ_M at 60% effort level, and extracted MAO. $ICC_{(2,k)}$ for muscle activation patterns ranged from 0.70 to 0.99, indicating high between-subject reliability from the plotted group mean. For all muscles, an increase in effort level did not significantly change asymmetry or Φ_M , indicating the roles of muscles did not change (aim 1).

Because of statistically symmetrical activation patterns, the vastus lateralis and the vastus medialis were classified as general joint stabilizers. All other muscles were statistically asymmetrical. RF, TFL, ST, ADD, and GM were classified as hip moment actuators because their Φ_M did not statistically differ from their reported hip MAO ($P > 0.05$). Similarly, ST, RF, and LG were classified as knee moment actuators because their Φ_M did not statistically differ from their reported knee MAO ($P > 0.05$). By contrast, MG and BF had Φ_M values significantly different from their reported knee and hip MAO ($P < 0.05$), classifying them as specific joint stabilizers.

Moments. $ICC_{(2,k)}$ values for internal net joint moment profiles ranged from 0.91 to 1.0, indicating very high between-subject reliability from the group mean depicted in Figure 4. In general, the required moment at each target location increased linearly with an increase in GRF effort. Sagittal plane moments were greatest at the hip with extension (group mean peak [GMP] = $0.898 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) magnitudes almost four times greater than knee flexion (GMP = $0.245 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) at posterior targets (hip extension $\Phi_T = 267.7^\circ$; knee flexion $\Phi_T = 270.6^\circ$). Hip flexion (GMP = $0.511 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) was approximately two times greater than knee extension (GMP = $0.299 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) at general anterior targets (hip

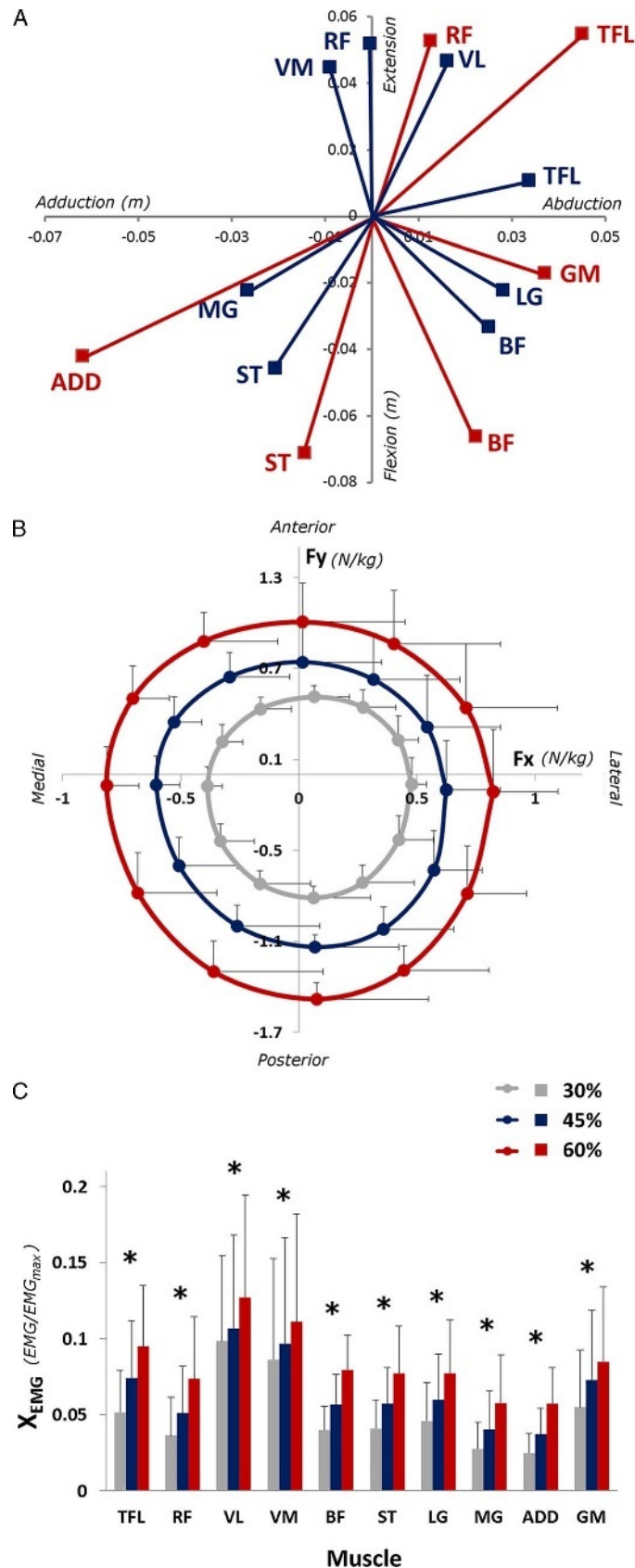


FIGURE 2—A, MAO of muscles crossing the knee (blue series) and hip (red series) relative to the given joints center of rotation when the knee is flexed to 25° and the hip is flexed to 28°. B, Group mean and SD anterior–posterior and medial–lateral force magnitudes required to reach each force target. C, Group mean and SD of X_{EMG} . *Significant difference in activation level ($P < 0.05$).

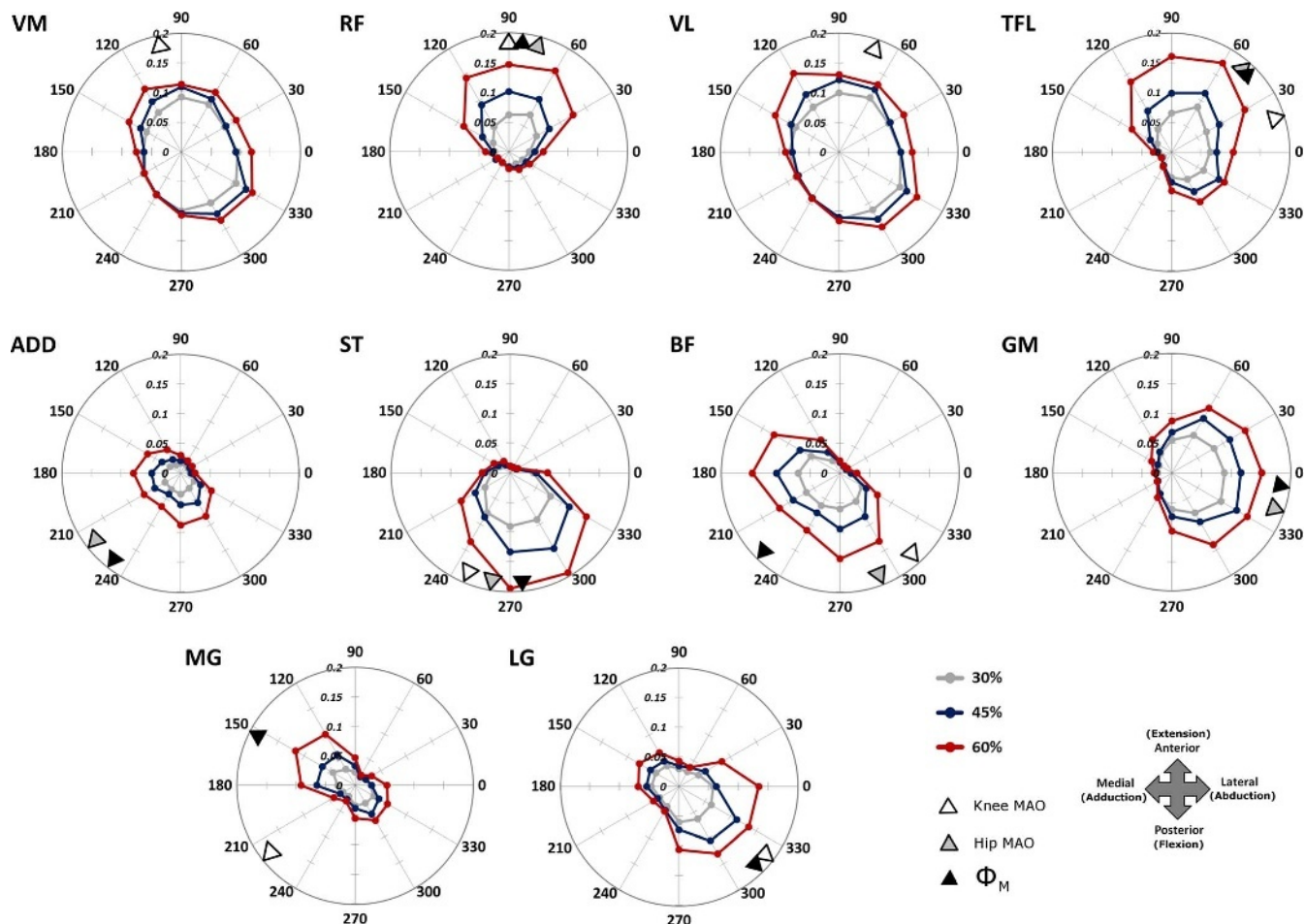


FIGURE 3—Group mean EMG polar plots at 30%, 45%, and 60% max effort level. Radii represent relative force loading direction, and where the pattern intersects, the radii is the normalized EMG amplitude (EMG/EMG_{max}) required to reach the force target. All plots are scaled to 0.2 EMG/EMG_{max} . Filled triangles along the circular trajectory indicate the mean direction of muscle activation (Φ_M) at the 60% effort level. Unfilled and gray triangles are the relative knee and hip MAO, respectively, as presented in Figure 2A. No triangle indicates the pattern was statistically symmetrical about its plot origin (no preferred activation direction).

flexion $\Phi_T = 81.7^\circ$; knee extension $\Phi_T = 84.6^\circ$). Similarly, hip adduction–abduction ($GMP = 0.462$ and $0.474 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) was approximately two times greater than knee adduction–abduction ($GMP = 0.218$ and $0.235 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) at general anterior–lateral (hip abduction $\Phi_T = 28.7^\circ$; knee abduction $\Phi_T = 39.0^\circ$) and posterior–medial (hip adduction $\Phi_T = 213.6^\circ$; knee adduction $\Phi_T = 218.0^\circ$) targets. Similar magnitudes of hip (internal $GMP = 0.218 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$; external $GMP = 0.202 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) and knee (internal $GMP = 0.208 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$; external $GMP = 0.180 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$) rotation moments were observed during anterior–medial (hip internal $\Phi_T = 164.6^\circ$; knee internal $\Phi_T = 146.3^\circ$), and posterior–medial targets (hip external $\Phi_T = 340.9^\circ$; knee external $\Phi_T = 320.8^\circ$). Subjects predominantly produced dorsiflexion moments at the ankle and coupled inversion with internal ankle rotation and eversion with external ankle rotation moments.

Regression models. All PLSR models showed significant associations between internal joint moments and muscle activation ($P < 0.0001$). However, the accuracy of prediction varied across muscles ($R^2 = 0.09$ – 0.58 , Fig. 5). The vastii had very weak prediction accuracy values ($R^2 < 0.10$),

with knee extension as the only significant positive predictor ($\beta = 0.115$ – 0.132). The LG and the MG had moderate prediction accuracy ($R^2 = 0.45$ – 0.47) with significant knee moment predictors of internal rotation for LG ($\beta = 0.20$) and external rotation for MG ($\beta = 0.25$). Hip muscles, ADD ($R^2 = 0.30$) and GM ($R^2 = 0.37$), had moderate prediction accuracies. Significant predictors for ADD were hip extension ($\beta = 0.28$) and hip adduction ($\beta = 0.17$). GM had hip abduction ($\beta = 0.18$) and internal hip rotation ($\beta = 0.23$) as significant predictors. RF ($R^2 = 0.58$) and TFL ($R^2 = 0.36$) had strong and moderate prediction accuracies with hip flexion (RF $\beta = 0.35$, TFL $\beta = 0.12$) and knee extension (RF $\beta = 0.34$, TFL $\beta = 0.12$) as significant predictors. TFL was also predicted by hip and knee abduction ($\beta_s = 0.08$). Last, hamstring muscles had strong prediction accuracies ($R^2 = 0.53$ – 0.58), with hip extension (BF $\beta = 0.10$, ST $\beta = 0.23$) and knee flexion (BF $\beta = 0.09$, ST $\beta = 0.11$) as significant predictors. ST was also predicted by internal hip ($\beta = 0.08$) and knee rotation ($\beta = 0.17$), whereas BF was also predicted by hip and knee adduction ($\beta = 0.12$) and hip and knee external rotation ($\beta = 0.06$ – 0.09).

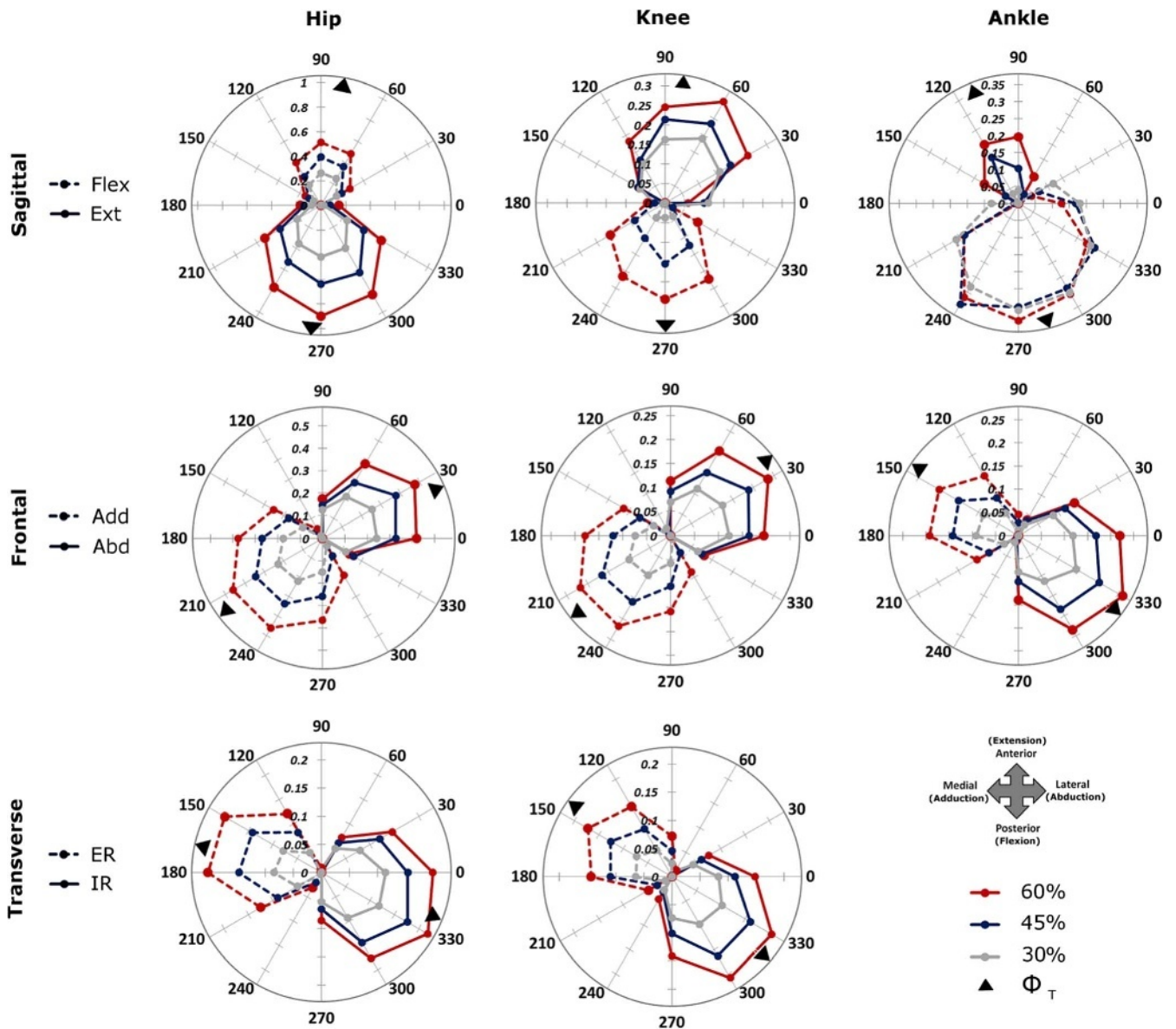


FIGURE 4—Group mean hip, knee, and ankle joint moment polar plots at 30%, 45%, and 60% effort level. Radii represent relative force loading direction, and where the pattern intersects, the radii is the normalized internal net joint moment ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$) required to reach the force target. Triangles along circular trajectory indicate the mean direction of the given moment (Φ_T) at the 60% effort level (ext, extension/plantarflexion; flex, flexion/dorsiflexion; add, adduction/inversion; abd, abduction/eversion; IR, internal rotation; ER, external rotation moments).

Table 1 summarizes the classification of muscle roles and the significant moment predictors (β) for each muscle.

DISCUSSION

Our study used a weight-bearing isometric GRF matching protocol to classify the roles of lower limb muscle as it relates to joint stability by quantifying the relationship between individual muscle activation patterns and internal net joint moments in all three loading planes (aim 1) and to determine whether the roles of muscles, as it relates to knee joint stability, changes with increasing force levels (aim 2). Our results in part support our hypotheses. First, the vastii yielded

symmetrical activation patterns and weak model-fit parameters classifying them to be general joint stabilizers. It was hypothesized that the activation of biarticular muscles would be predicted by one general moment at each articulating joint; however, biarticular muscles had more than one significant moment predictor indicating their roles to be multifactorial. Second, an increase in effort level did not significantly change Φ_M or asymmetry of muscle activation patterns indicating the roles of muscles did not change with an increase in force level.

To date, most research on dynamic knee joint stability has focused on how knee muscles activate as a function of their anatomical orientation (4,11) and typically classify activation

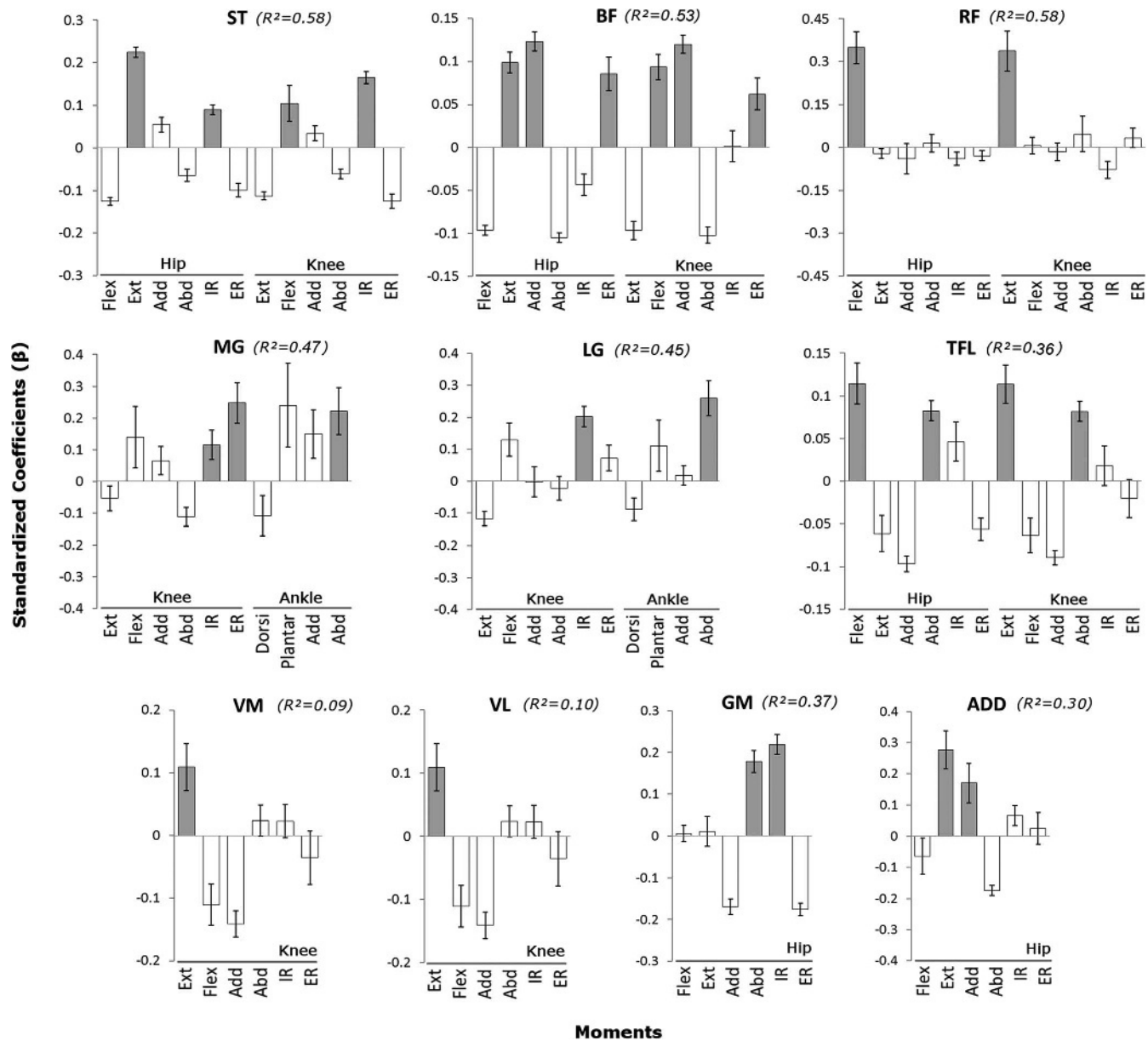


FIGURE 5—Standardized coefficients (β) of moments for bi- and monoarticular muscles model goodness of fit indicated with R . Only significant positive β coefficients were considered meaningful. These are represented as filled data points. Error bars indicate SE of β . Only joints over which the given muscle crossed were included in the comparisons (ext, extension; flex, flexion; add, adduction; abd, abduction; IR, internal rotation; ER, external rotation, dorsi, dorsiflexion, plantar, plantarflexion moments).

as either agonistic or antagonistic to sagittal plane motion (3,19). The interpretation of muscle roles is also often described from non-weight-bearing tasks which limits the functional relevance of the activations (3,11,38). Literature pertaining to the muscular contributions to frontal and transverse plane loads, in particular during weight bearing, is therefore limited. Previously, we proposed knee definitions for the roles of knee joint muscles as it relates to stability. Our analysis is based on the modulation of muscle activation across various force directions and how this relates to MAO (16,17,37). The addition of PLSR to our methods has further elucidated the direct relationship between individual muscle

activity and internal joint moments. For example, based on its activation pattern and MAO, the ST was classified as a moment actuator for knee flexion. Our PLSR expanded this role to include a specific stabilizing role for knee rotation because ST activation was also significantly predicted by internal knee rotation moments.

To generate GRF, subjects elicited various combinations of internal flexion–extension, adduction–abduction, and rotational moments at the hip, knee, and ankle joints. Both the moment and muscle activation profiles were extremely similar across subjects (EMG ICCs > 0.7; moment ICCs > 0.9), suggesting a fundamental strategy to coordinate and distribute

TABLE 1. Summary of variables used to classify roles of muscles acting at the hip and knee joints.

Muscle	Asymmetrical?	Does MAO = Φ_M ?		Role Classification		Prediction Accuracy	Significant Predictors (β)	
		Hip	Knee	Hip	Knee		Hip	Knee
BF	Yes	No	No	SJS	SJS	Strong	Ext, Add, IR	Flex, Add, IR
ST	Yes	Yes	Yes	MA	MA	Strong	Ext, IR	Flex, IR
TFL	Yes	Yes	No	MA	SJS	Moderate	Flex, Abd	Ext, Abd
RF	Yes	Yes	Yes	MA	MA	Strong	Flex	Ext
VL	No		–		GJS	Very weak		Ext
VM	No		–		GJS	Very weak		Ext
LG	Yes		Yes		MA	Moderate		IR
MG	Yes		No		SJS	Moderate		ER
ADD	Yes	Yes		MA		Moderate	Ext, Add	
GM	Yes	Yes		MA		Moderate	Abd, IR	

Φ_M , mean direction of muscle activation; GJS, general joint stabilizer; SJS, specific joint stabilizer; MA, moment actuator; Ext, extension; Flex, flexion; Add, adduction; Abd, abduction; IR, internal rotation; ER, external rotation moments; VM, vastus medialis; VL, vastus lateralis.

lower limb net joint moments to apply a given GRF at the ground–foot interface. Unlike previous works, we were able to provide a quantifiable relationship between individual muscle EMG and lower limb internal net joint moments during weight bearing, and thus we have the ability to isolate the functional roles of muscles.

Quadriceps muscles. The quadriceps muscles undoubtedly function as knee extensors, and because of their MAO, activation will generate an extensor moment. Yet this does not fully describe their functional role. Because of symmetrical muscle activation patterns, we classified the vastii as general joint stabilizers. This classification was also supported by the very weak prediction accuracy values ($R^2 < 0.1$), indicating a change in internal joint moments will unlikely result in a change in muscle activation. By contrast, RF had an asymmetrical activation with its Φ_M consistent with its MAO and significant model predictors. As such, the RF was classified as a moment actuator for hip flexion and knee extension moments. The different roles between vastii and RF demonstrate the complexity for force control in weight bearing. van Ingen Schenau et al. (39) suggested that perhaps mono- and biarticular muscles play different roles in lower limb force control: the monoarticular muscles provide a general gross force, whereas biarticular muscles “fine tune” the distribution of the net joint moments across adjacent joints. Considering the vastii are 1) primary contributors to axial knee joint force (36), 2) can provide the greatest contribution to knee joint rotational stiffness in all three flexion–extension, varus–valgus, and transverse loading axes (13), and 3) they actually have a protective effect on the anterior cruciate ligament (ACL) when physiological loading magnitudes are applied *in vitro* (22). We postulate that vastii muscles contract to increase compressive forces, essentially bracing the knee, so biarticular hip muscles, such as the RF, can generate, and transmit the moments needed to direct the GRF at the foot–ground interface.

Hamstrings muscles. The role of the hamstrings as it relates to lower limb function is often associated with generating knee flexion moments. Again, anatomically if one focuses on the knee and their MAO this is a correct observation. Because ST's knee MAO was not statistically different from its activation pattern's Φ_M , we first assumed that it was the primary contributor to knee flexion. This is supported by the ST's PLSR model. However, the β for hip

extension is nearly 2.5x greater than knee flexion. By contrast, the BF's knee MAO was statistically different from its Φ_M . Accordingly, it was classified as a specific joint stabilizer; but the PLSR model determined that knee flexion was also significantly associated with BF activity. Similar to the ST, the BF's β for hip extension is greater than knee flexion. Considering dominant hip moment strategies were elicited for generating posterior GRF (hip extension nearly four times greater than knee flexion), we may assume that the hamstrings function more as hip extensors. Similarly, in a contact force control task involving all three lower limb joints, van Deursen et al. (38) reported that the activations of the biarticular RF and hamstring muscles had the highest correlations with changes in hip joint moments compared with knee joint moments.

In weight-bearing conditions, the activation of the hamstrings is commonly interpreted as antagonist activation against the quadriceps to reduce knee extension moments and prevent anterior tibial translation (29). However, moment arms of the hamstrings also possess substantial rotational components (1,2,24). Our results also show a relatively high association of ST and BF with internal and external knee rotation, respectively, suggesting that hamstring activation is essential for stabilizing the knee against torsional loads. In fact, extension of the knee using isolated quadriceps force causes an internal rotation of the tibia relative to the femur (40)—which is linked to the ACL injury mechanism (7). This internal rotation is reduced only when hamstring loads, especially the BF, are added resulting in knee motion with a neutral alignment (40).

Interestingly, BF was also associated with hip and knee adduction moments. This may be from one of two scenarios: 1) it is generating an antagonistic force to oppose hip and knee adduction moments or 2) its activation profile reflects a bimodal pattern encompassing external hip/knee rotation and hip extension/knee flexion. For the latter, when combined, the mean principle moment directions (Φ_T) for external hip/knee rotation and hip extension/knee flexion is equal to 211°, which is extremely close to the principle directions of hip ($\Phi_T = 213^\circ$) and knee adduction ($\Phi_T = 218^\circ$).

Gastrocnemius muscles. LG and MG also demonstrated significant associations with respective internal and external knee rotation moments. In addition to its role of a

knee joint flexor, the gastrocnemius has the potential to contribute to rotational knee moments (2) and is commonly the first to activate during rotational perturbations (12). Despite this and its changing activation after ACL injury (35), little literature is available about its contribution at the knee, possibly because of its confounding role at the ankle. On the basis of its anatomical alignment, gastrocnemius activity has been hypothesized to create anterior shear force on the tibia, resulting in an increase in ACL loading (33). Several studies have tested this hypothesis: Durselen et al. (15) applied a 550-N gastrocnemius load *in vitro*, and there was no effect on ACL strain at any flexion angles; Fleming et al. (18) electrically stimulated the gastrocnemius muscle to produce a plantarflexion moment, which increased *in vivo* ACL strain between 0° and 30° knee flexion; Morgan et al. (32) estimated *in silico* that elevated gastrocnemius activity is synergistic with the quadriceps to help compress the knee and lower ACL forces during single-leg jump landing. Despite contradictory evidence, reported differences in gastrocnemius activity exist in populations with knee instability (35), and this changing activation will alter knee loads. It remains unclear whether gastrocnemius activity as it relates to knee joint stability is protective or not and further investigation is warranted.

Hip joint muscles. Because gluteal muscles, hip abductors, and hip adductors are large determinants of femoral orientation and knee joint loads (36), this study analyzed their contribution to hip moments. As expected, the ADD muscles and GM were predicted well with hip adduction and abduction moments, respectively. Because of the difficulty in isolating a single ADD muscle with surface EMG, our ADD EMG signal is considered a summation of inner thigh muscle activity. This could explain why its activation was also predicted by hip extension. Internal hip rotation was also deemed significant predictors for GM, but we postulate it is antagonistically activating to stabilize the hip. This is a noteworthy observation because reduced hip function destabilizes the femur, causing an increase in frontal plane motion and knee joint moments (34).

Last, although the TFL is generally considered a hip joint muscle, we included knee moments in its model because it attaches to the iliotibial tract and inserts on the lateral condyle of the tibia (2,42). Results indicate the TFL contributes to hip flexion and knee extension. Increased knee abduction loads are linked to the ACL injury mechanism (7), so it is surprising that TFL was also the only muscle to have knee abduction moments as a significant predictor. Previous works found that the TFL contributed up to 25% of muscular loads to lateral knee compartment during gait (42) and assisted in knee extension and minor valgus moment generation (11). Like the gastrocnemius muscles, the role of the TFL as it relates to knee joint stability is not well understood and also warrants future examination.

Our protocol was conducted while weight bearing, which we believe to be a functionally relevant task (consider bracing yourself on a moving bus, opening a heavy door, or preparing to rise from a seat). However, it is important to note that the

EMG–moment relationships observed in the present study and how it relates to those observed in ballistic activities remain unclear. Considering most existing studies associate a proportional increase activation to a proportional increase in joint moments based on relationships established from non–weight-bearing and/or uniaxial *in vivo* studies (6,38), *in silico* investigations that may be prone to poor predictions of agonist and antagonist activations (23), or *in vitro* anatomical investigations based on muscle lines of action and moment arms (29,44), we believe our results provide greater and novel insight into the functional role of the investigated muscles. Furthermore, the moment arms extracted from the musculoskeletal model (21) are not subject specific. Because significant between-subject differences exist in anatomical muscle paths (8), the extracted moment arms may not be representative of every participant. Last, the classification of muscle roles as it relates to knee joint stability is associated with only one bodily configuration. In particular, for the biarticular muscles, muscle activation patterns are subject to change depending on flexion angle (11). We opted for this position because the hip and knee flexion angles are common to several sporting maneuvers associated with ACL injury, such as side cuts, braking motions, and landing (7). We wanted to elicit as much muscular protection as possible, and shallower knee flexion angle would have increased the reliance on the soft tissues and reduced mechanical advantages of the knee muscles (27). Nevertheless, similar activation patterns in a standing force matching have been reported with extended knees (28).

In summary, our results show that activation is not always dependent on anatomical orientation. Using PLSR models, we quantitatively associated internal moment generation with changes in individual muscle activations, thus specifying the functional contributions of muscles to maintaining knee joint stability. This can be particularly useful for delineating the roles of the biarticular knee joint muscles. Moments required to modulate GRF are primarily generated by the biarticular muscles crossing the hip, whereas uniaxial muscles crossing the knee are responsible for producing compressive forces, essentially bracing the knee, so that proximal hip loads can be transmitted down to the foot–ground interface. Our results also emphasize the importance of the hamstrings and gastrocnemius muscles in supporting rotational torques at the knee. Functional roles of lower limb muscles need to be reconsidered when describing how muscle activity relates to moment generation, which could have implications in robotics, musculoskeletal modeling, sports sciences, and rehabilitation.

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REFERENCES

- Aalbersberg S, Kingma I, Ronsky JL, Frayne R, van Dieën JH. Orientation of tendons in vivo with active and passive knee muscles. *J Biomech.* 2005;38(9):1780–8.
- Agur AMR, Dalley AF. *Grant's Atlas of Anatomy.* Baltimore (MA): Lippincott Williams & Wilkins; 2005.
- Alkjær T, Simonsen EB, Magnusson SP, Dyhre-Poulsen P, Aagaard P. Antagonist muscle moment is increased in ACL deficient subjects during maximal dynamic knee extension. *Knee.* 2012;19(5):633–9.
- Andriacchi TP, Andersson GB, Ortengren R, Mikosz RP. A study of factors influencing muscle activity about the knee joint. *J Orthop Res.* 1984;1(3):266–75.
- Berens P. CircStat: a MATLAB toolbox for circular statistics. *J Stat Softw.* 2009;31(10):1–20.
- Bigland-Ritchie B. EMG/force relations and fatigue of human voluntary contractions. *Exerc Sport Sci Rev.* 1981;9:75–117.
- Boden BP, Dean GS, Feagin JA Jr, Garrett WE Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* 2000;23(6):573–8.
- Brand RA, Crowninshield RD, Wittstock CE, Pedersen DR, Clark CR, Van Krieken FM. A model of lower extremity muscular anatomy. *J Biomech Eng.* 1982;104(4):304–10.
- Brand RA, Pedersen DR, Friederich JA. The sensitivity of muscle force predictions to changes in physiologic cross-sectional area. *J Biomech.* 1986;19(8):589–96.
- Buchanan TS, Almdale DP, Lewis JL, Rymer WZ. Characteristics of synergic relations during isometric contractions of human elbow muscles. *J Neurophysiol.* 1986;56(5):1225–41.
- Buchanan TS, Lloyd DG. Muscle activation at the human knee during isometric flexion–extension and varus–valgus loads. *J Orthop Res.* 1997;15(1):1–17.
- Carcia CR, Shultz SJ, Granata KP, Perrin DH, Martin RL. Females recruit quadriceps faster than males at multiple knee flexion angles following a weight-bearing rotary perturbation. *Clin J Sport Med.* 2005;15(3):167–71.
- Cashaback JG, Potvin JR. Knee muscle contributions to joint rotational stiffness. *Hum Mov Sci.* 2012;31(1):118–28.
- Delp SL, Anderson FC, Arnold AS, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng.* 2007;54(11):1940–50.
- Dürselen L, Claes L, Kiefer H. The influence of muscle forces and external loads on cruciate ligament strain. *Am J Sports Med.* 1995;23(1):129–36.
- Flaxman TE, Speirs AD, Benoit DL. Joint stabilisers or moment actuators: the role of knee joint muscles while weight-bearing. *J Biomech.* 2012;45(15):2570–6.
- Flaxman TE, Smith AJ, Benoit DL. Sex-related differences in neuromuscular control: implications for injury mechanisms or healthy stabilisation strategies? *J Orthop Res.* 2014;32(2):310–7.
- Fleming BC, Renstrom PA, Ohlen G, et al. The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. *J Orthop Res.* 2001;19(6):1178–84.
- Granata KP, Wilson SE, Padua DA. Gender differences in active musculoskeletal stiffness. Part I. Quantification in controlled measurements of knee joint dynamics. *J Electromyogr Kinesiol.* 2002;12(2):119–26.
- Haenlein M, Kaplan AM. A beginner's guide to partial least squares analysis. *Underst Stat.* 2004;3(4):283–97.
- Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech.* 2010;43(14):2709–16.
- Hashemi J, Chandrashekar N, Jang T, Karpal F, Oseto M, Ekwaro-Osire S. An alternative mechanism of non-contact anterior cruciate ligament injury during jump-landing: in-vitro simulation. *Exp Mech.* 2007;47(3):347–54.
- Herzog W, Binding P. Predictions of antagonistic muscular activity using nonlinear optimization. *Math Biosci.* 1992;111(2):217–29.
- Herzog W, Read LJ. Lines of action and moment arms of the major force-carrying structures crossing the human knee joint. *J Anat.* 1993;182(Pt 2):213–30.
- Krishnan C, Williams GN. Sex differences in quadriceps and hamstrings EMG–moment relationships. *Med Sci Sports Exerc.* 2009;41(8):1652–60.
- Levin O, Wenderoth N, Steyvers M, Swinnen SP. Directional invariance during loading-related modulations of muscle activity: evidence for motor equivalence. *Exp Brain Res.* 2003;148(1):62–76.
- Lloyd DG, Buchanan TS. A model of load sharing between muscles and soft tissues at the human knee during static tasks. *J Biomech Eng.* 1996;118(3):367–76.
- MacLeod TD, Manal K, Silbernagel KG, Snyder-Mackler L, Buchanan TS. Characteristics of human knee muscle coordination during isometric contractions in a standing posture: the effect of limb task. *J Electromyogr Kinesiol.* 2013;23(6):1398–405.
- MacWilliams BA, Wilson DR, Desjardins JD, Romero J, Chao EY. Hamstrings cocontraction reduces internal rotation, anterior translation, and anterior cruciate ligament load in weight-bearing flexion. *J Orthop Res.* 1999;17(6):817–22.
- Mantovani G, Lamontagne M. How different marker sets affect joint angles in inverse kinematics framework. *J Biomech Eng.* 2016 [Epub ahead of print].
- McGraw KO, Wong SP. Forming inferences about some intraclass correlation coefficients. *Psychol Methods.* 1996;1(1):30–46.
- Morgan KD, Donnelly CJ, Reinbolt JA. Elevated gastrocnemius forces compensate for decreased hamstrings forces during the weight-acceptance phase of single-leg jump landing: implications for anterior cruciate ligament injury risk. *J Biomech.* 2014; 47(13):3295–302.
- O'Connor JJ. Can muscle co-contraction protect knee ligaments after injury or repair? *J Bone Joint Surg Br.* 1993;75(1):41–8.
- Powers CM. The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. *J Orthop Sports Phys Ther.* 2010;40(2):42–51.
- Rudolph KS, Axe MJ, Buchanan TS, Scholz JP, Snyder-Mackler L. Dynamic stability in the anterior cruciate ligament deficient knee. *Knee Surg Sports Traumatol Arthrosc.* 2001;9(2):62–71.
- Sasaki K, Neptune RR. Individual muscle contributions to the axial knee joint contact force during normal walking. *J Biomech.* 2010;43(14):2780–4.
- Smith AJ, Flaxman TE, Speirs AD, Benoit DL. Reliability of knee joint muscle activity during weight bearing force control. *J Electromyogr Kinesiol.* 2012;22(6):914–22.
- Van Deursen RWM, Cavanagh PR, van Ingen Schenau GJ, Becker MB, Ulbrecht JS. The role of cutaneous information in a contact control task of the leg in humans. *Hum Mov Sci.* 1998;17(1):95–120.
- van Ingen Schenau GJ, Boots PJ, de Groot G, Snackers RJ, van Woensel WW. The constrained control of force and position in multi-joint movements. *Neuroscience.* 1992;46(1):197–207.
- Victor J, Labey L, Wong P, Innocenti B, Bellemans J. The influence of muscle load on tibiofemoral knee kinematics. *J Orthop Res.* 2010;28(4):419–28.
- Wilkie DR. The relation between force and velocity in human muscle. *J Physiol.* 1949;110(3–4):249–80.
- Winby CR, Lloyd DG, Besier TF, Kirk TB. Muscle and external load contribution to knee joint contact loads during normal gait. *J Biomech.* 2009;42(14):2294–300.
- Wold S, Ruhe A, Wold H, Dunn IW. The Collinearity Problem in Linear Regression. The Partial Least Squares (PLS) Approach to Generalized Inverses. *SIAM J Sci Stat Comput.* 1984;5(3):735–43.
- Zheng L, Li K, Shetye S, Zhang X. Integrating dynamic stereo-radiography and surface-based motion data for subject-specific musculoskeletal dynamic modeling. *J Biomech.* 2014;47(12): 3217–21.