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Joint stabilisers or moment actuators: The role of knee joint muscles while weight-bearing

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ABSTRACT

Previous investigations have identified the roles of knee joint muscles in supporting external loads during non-weight-bearing tasks and found these to depend on moment arm orientation (MAO). However, during weight-bearing tasks ground reaction forces (GRF) are transferred up through the knee, subjecting it to large multi-directional forces and stability is dependent on articular geometry, loading, and muscle activation. The purpose of this study was to investigate activation strategies used by healthy individuals to generate and support highly controlled GRF during weight-bearing.

Twenty healthy males (23.9 ± 1.9 yrs) stood with their foot in a boot fixed to a force platform. Subjects controlled an onscreen cursor by modulating normalised GRF and were required to produce 30% of their maximal force in 12 directions of the horizontal plane while maintaining 50% body weight on the test leg. Lower limb electromyography, kinematics and kinetics were recorded for each trial. Mean muscle activation was plotted in polar coordinates based on GRF orientation. Muscle activation symmetry was determined and when applicable, the mean direction of activation and muscle specificity index reported.

The measured GRF were comparable to activities of daily living (0.48–0.58 ± 0.17–0.19 N/kg in horizontal plane). Muscle activations were repeatable (ICCs: 0.78–0.98), however, only semitendinosus (ST) activation was indicated by its MAO. Considering the joint moments and activations patterns we therefore classified muscles as: (1) general joint stabilisers (vastus lateralis and medialis), (2) specific joint stabiliser (BF), and (3) moment actuators (ST and rectus femoris). General joint stabilisers were active in all load directions; specific stabilisers were active in directions opposite their MAO; moment actuators had higher specificities and activations corresponding to their MAO. We suggest the stabiliser muscles create a rigid mechanical linkage at the knee which allows the actuators of the hip and knee to modulate GRF.

1. Introduction

The knee is subjected to large forces during activities of daily living (ADLs). The ability of the knee to efficiently support these loads results from the integration of articular geometry, soft tissue restraints, muscle action, and body mass (BM). Of these, muscles are the only active regulators of load distribution; thus, neuromuscular function is a crucial determinant of knee joint stability (Hsieh and Walker, 1976).

Evaluations of neuromuscular function with respect to knee joint stabilisation typically involve dynamic task assessments (e.g. walking, jumping, perturbations, etc.) (da Fonseca et al., 2006; Hanson et al., 2008; Hortobagyi and DeVita, 2000), but isolating muscle function by uncoupling biomechanical and neuromuscular contributions to force generation is difficult in an anatomically complex system (Wilkie, 1950). It may be beneficial to limit the biomechanical factors so roles of individual muscles can be more clearly associated with the measured task. To achieve this, isometric knee exercises have been used, however, evaluations usually focused on flexion and extension (Hortobagyi et al., 2004; Rainoldi et al., 2001) even though functional activities do not limit loads to these directions. Appreciating this, Buchanan and Lloyd (1997), and many studies since (Benoit et al., 2006; Krishnan et al., 2008; Lloyd and Buchanan, 1996, 2001; Williams et al., 2003) implemented an isometric force matching protocol to evaluate neuromuscular strategies used to support various combinations of flexion-extension-varus-valgus loads at the knee. Subsequently, electromyographic (EMG) data were plotted in polar coordinates (Dewald et al., 1995) so muscle activation patterns could be quantified, described, and compared across test populations. These studies have shed light on the roles of knee joint muscles in supporting direction-dependent loads, revealing that muscle activation is directly related to its moment arm orientation (MAO) (defined as the relative position

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of the muscle's line of action from the joint's centre of rotation), however, subjects were seated and the foot unrestricted.

Since most functional activities are performed while weight-bearing, which alters lower limb kinematics, joint loading conditions, and corresponding muscle activation characteristics (Escamilla et al., 1998; Kiefer et al., 1998; Shultz et al., 2009; Wilk et al., 1996) extending the results of these studies to functional tasks may be limited. We have therefore developed an approach that combines the benefits of isometric, direction-dependent force matching with the physiological influences of bearing weight. The purpose of this study was to investigate muscle activation strategies used by healthy individuals to generate and support highly controlled GRF during weight-bearing. We hypothesise that muscle activation patterns will be dictated by knee joint stabilisation strategies rather than MAO.

2. Methods

2.1. Subjects

Twenty healthy active young males (age$=$23.9 $\pm$ 1.9 yrs; weight$=$79.7 $\pm$ 9.3 kg; height$=$1.77 $\pm$ 0.05 m; body mass index$=$25.4 $\pm$ 0.06 kg/m^2) participated in this study. Exclusion criteria were previous reports of significant lower limb injuries (e.g. ligament rupture), lower limb sprains or fractures within 6 months of participation, or any other physical impairment that may influence knee function. All subjects read and signed an informed consent form. The study was approved by the University of Ottawa Research Ethics Board.

2.2. Protocol

Bipolar surface EMG electrodes (DS-B04, Bagnoli-16, Delsys Inc., Boston, MA) were placed over the muscle bellies of the rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), semitendinosus (ST), biceps femoris (BF), medial gastrocnemius (MG), lateral gastrocnemius (LG), and tensor fascia lata (TFL) of each leg following the recommendations by SENIAM (Hermens et al., 2000) and DeLuca (1997). Anthropometric measurements were recorded and a 5 min warm-up on a stationary bike (Monark AB, Sweden; 90 RPM with no resistance) was performed. Maximal voluntary isometric contraction (MVIC) data were then collected using an isokinetic dynamometer (850-000, Biodex, New York, USA) and analysed with custom made software (Labview 8.20; National Instruments Corp., Austin, TX, USA). Plantar flexion, knee extension, and knee flexion were recorded at relaxed hip, knee, and ankle joint angles of 30°, 30°, and 10°, respectively. Hip abduction was recorded in a standing position. With verbal encouragement, subjects were given 10 s to scale up to their maximal effort and hold it for approximately 3 s. Subjects then had 45 reflective markers placed on various landmarks according to a modified Plug-in Gait marker set (Vicon, Oxford Metrics, Oxford, UK; Beaulieu et al., 2010).

A novel, reliable (Smith et al., in press) weight-bearing force matching protocol was then used to assess a subject's voluntary muscle activation patterns while isometrically modulating GRF. Following an introduction and practice trials, subjects stood with their dominant foot, defined by the leg used to kick a ball, in a water-ski boot (Bio, O'Brien, Redmond, WA, USA) attached to a force platform (FP4060-08, Bertec Corporation, Columbus, OH, USA). Subjects were encouraged to maintain 30° hip flexion, 30° knee flexion, and 10° ankle flexion during testing. A projected image of a cursor and a target were placed in front of the subject, providing visual feedback of the direction and magnitude of GRF applied to the force platform (Fig. 1). GRF controlled a cursor that moved with three degrees-of-freedom: (1) anterior/posterior loads ($\pm Fy$-axis) moved the cursor upward/downward, (2) medial/lateral loads ($\pm Fx$-axis) to the left/right, and (3) inferior/superior loads ($\pm Fz$-axis) made the cursor smaller/larger (i.e. controlling the percent BM applied to the ground). Maximal GRF were then collected; subjects maintained equal BM on each leg and exerted a maximal effort against the force platform in each of the anterior ($+Fy$), posterior ($-Fy$), medial ($-Fx$), and lateral ($+Fx$) loading directions.

Targets were then scaled for testing to 30% $\pm$ 3.9% of the recorded maximal GRF in the horizontal plane, and 50% $\pm$ 5% BM along Fz. A successful target match consisted of holding the cursor over the target (thus applying the desired GRF magnitude and direction) for 1 s, which triggered data collection (see below). Three repetitions of 12 targets spaced by 30° about a circular trajectory appeared in a random order. Subjects were given a minimum of 30 s rest (more if required) before attempting the next target.

2.3. Data collection and processing

The 1 s of EMG, GRF and kinematic data corresponding to the successful GRF matching period was recorded for each target using a digitally synchronised *Matlab* application (EMG; Matlab 2007b, Mathworks, Natick, Massachusetts) and a nine-camera infrared motion analysis system with *Vicon Nexus* (GRF and kinematics; *Nexus* version 1.3, Vicon MX-13, Oxford Metrics, Oxford, UK). GRF and kinematic trajectories were sampled at 1000 Hz and 200 Hz, respectively, and filtered with a 4th order Butterworth dual low-pass (20 Hz) filter (Bisseling and Hof, 2006). A modified Plug-in-Gait model (Beaulieu et al., 2010) calculated the hip, knee, and ankle joint angles and moments with *Nexus* software.

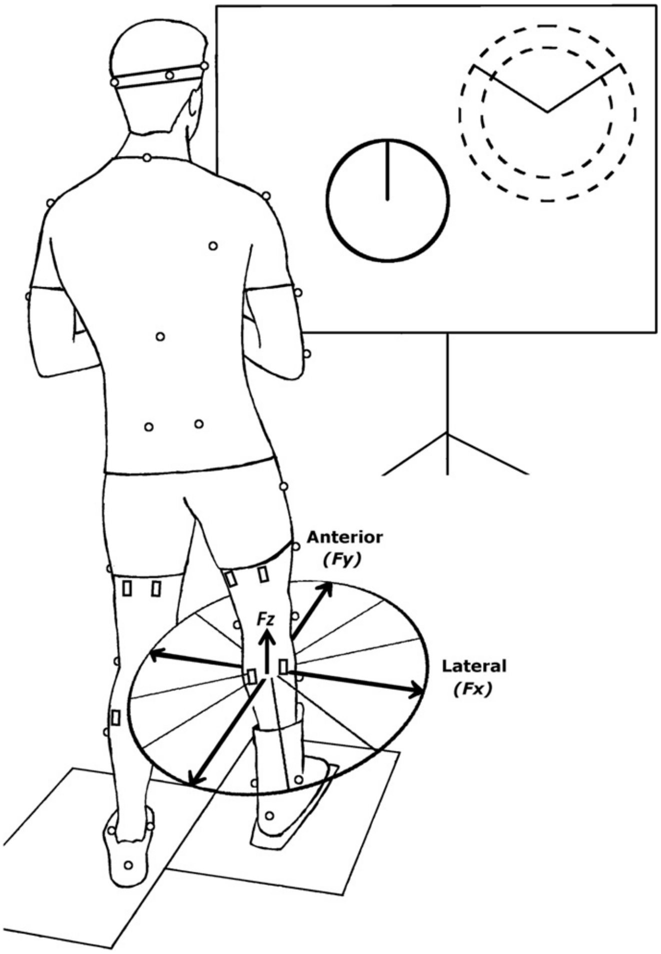

Fig. 1. Laboratory setup: the subject stands with their dominant foot in a boot fixed to a force platform and the opposite foot located posteriorly and adjacently. A projector displays biofeedback of the GRF generated by the subject as a cursor (solid circle). Subjects must position the cursor between the target's two rings (dashed lines) by modulating the GRF in the horizontal plane (Fx and Fy), with Fx and Fy corresponding to medial and anterior cursor movements, respectively. Cursor diameter increases or decreases with more or less body weight loading (Fz), respectively. A successful match requires 30% of a pre-recorded maximal effort in the horizontal plane while maintaining 50% body weight on the test foot.

EMG data were sampled at 1000 Hz, band-pass filtered at 20–450 Hz using a 16-bit A/D conversion board (NI PCI 6229, National Instruments Corp., Austin, TX), then bias corrected, full wave rectified and filtered (4th order Butterworth dual low-pass (6 Hz) filter). The test leg's target match EMG data was normalised to percent MVIC and ensemble averaged across repetitions, corresponding to an EMG vector (EMGi)—its orientation and magnitude representing GRF direction and percent MVIC, respectively. EMGi vectors were plotted in polar coordinates to visually represent muscle activation patterns (Dewald et al., 1995).

To determine if subject's activation magnitudes modulated across targets in a similar manner, percent EMG was scaled to maximum value recorded in the given muscle of each subject. Then, between-subject reliability of activation profiles was evaluated with intra-class correlation (ICC$_{(2,\,k)}$) coefficients (Portney and Watkins, 2000; Smith et al., in press) in SPSS (v18.0, IBM, Armonk, NY).

To quantitatively describe these patterns, we first tested for asymmetry about the polar plot origin following methods used by Curray (1956):

$$p = e^{-L^2 n} 10^{-4}$$

where p is the probability of observing non-random asymmetry, e is the base of natural logarithm, L is the mean vector magnitude, and n is the number of observations. Asymmetry was observed if $p < 0.05$, indicating greater muscle activation in one GRF direction relative to another. Symmetrical activation (similar EMG levels in all directions) was observed if $p > 0.05$.

The "mean direction of muscle activity" (Φ) was determined for muscles with asymmetrical activation by taking the arc tan of the vectors' summed Cartesian coordinates (x_i and y_i)

$$\Phi = \tan^{-1}\left(\frac{\sum y_i}{\sum x_i}\right)$$

In addition, the variance of muscle activation about the Φ was described as a "specificity index" (SI), derived from the ratio of the muscle's actual resultant vector to its absolute resultant vector (Dewald et al., 1995; Williams et al., 2003)

$$\mathrm{SI} = \frac{\left[\left(\frac{\sum x_i}{n}\right)^2 + \left(\frac{\sum y_i}{n}\right)^2\right]^{-1/2}}{\left[\left(\frac{\sum |x_i|}{n}\right)^2 + \left(\frac{\sum |y_i|}{n}\right)^2\right]^{-1/2}}$$

The SI ranged between 0.0 and 1: 0.0 indicates a muscle was equally active in all target directions (completely non-specific; depicted by a circular activity pattern—symmetrical about the polar plot origin); whereas 1.0 indicates a muscle was only active at one target (completely specific; activity pattern depicted by a single radius).

Finally, a "mean magnitude of muscle activity" (X_{EMG}) was computed for all muscles by averaging all the normalised EMGi at every target location

$$X_{EMG} = \frac{\sum \mathrm{EMGi}}{n}$$

3. Results

3.1. Ground reaction forces

The mean (standard deviation) normalised forces expressed in newtons per kilogram of BM required to match a target were 0.48 ($\pm$0.17), 0.58 ($\pm$0.18), 0.49 ($\pm$0.17), and 0.52 ($\pm$0.19) N/kg in the lateral (0°), anterior (90°), medial (180°), and posterior (270°) loading directions, respectively.

3.2. Joint angles and moments

Hip, knee, and ankle flexion mean (standard deviation) angles were 25.05° (7.39°), 23.12° (5.20°), and 6.13° (5.61) across all subjects and trials. All were normally distributed (Shapiro-Wilk test, $p > 0.05$; SPSS v18.0).

Group mean lower limb joint moments are expressed as newton metres per kilogram of BM (Nm/kg) (Fig. 2). Generally, hip joint moments were greater than knee and ankle joint moments in all functional axes. For moments about the x-axis, knee extension (0.15–2.41 Nm/kg) was greatest at targets with anterior components. Conversely, minimal knee flexion moments (<0.05 Nm/kg) were present at posteriorly located targets while relatively large hip extension moments (0.47–0.53 Nm/kg) were noted. Knee and hip joint moments were greatest about the y-axis; adduction moments peaked (hip=0.62 and knee=0.34 Nm/kg) at targets with medial components, and modulated to minimal levels (<0.05 Nm/kg) at laterally located targets. External rotation was observed at all joints during anterior, posterior, and medial loading, and peaked when medial and anterior loading components were combined (90–180°).

3.3. Muscle activation patterns

ICCs for mean activation patterns (Fig. 3) ranged from 0.78 to 0.98 (Table 1) indicating high levels of between-subject reliability (Portney and Watkins, 2000).

The remaining EMG variables (Φs, SIs, and X_{EMG}s) are presented in Fig. 4A, B, and C, respectively. Five (RF, BF, ST, MG, and TFL) of eight muscles had asymmetrical activation patterns ($p < 0.01$). ST and MG demonstrated the highest specificity (SIs=0.43 and 0.42) with Φ directed in the posterior (255°) and anterior-medial (146°) loading directions, respectively. BF and TFL's activation patterns were also asymmetrical but lower SIs were observed (BF=0.36; TFL=0.35). Their Φ occurred in the general medial (177°) and lateral (340°) loading directions, respectively. RF had the lowest specificity (SI=0.26) about its Φ at 84°. VL, VM, and LG displayed symmetrical patterns ($p > 0.05$), indicating similar activation magnitudes occurred across all loading directions. VL, VM, and BF had the greatest X_{EMG} values ($>20\%$ MVIC) while relatively smaller X_{EMG} levels ($<11\%$ MVIC) were demonstrated in RF, ST, LG, MG and TFL muscles.

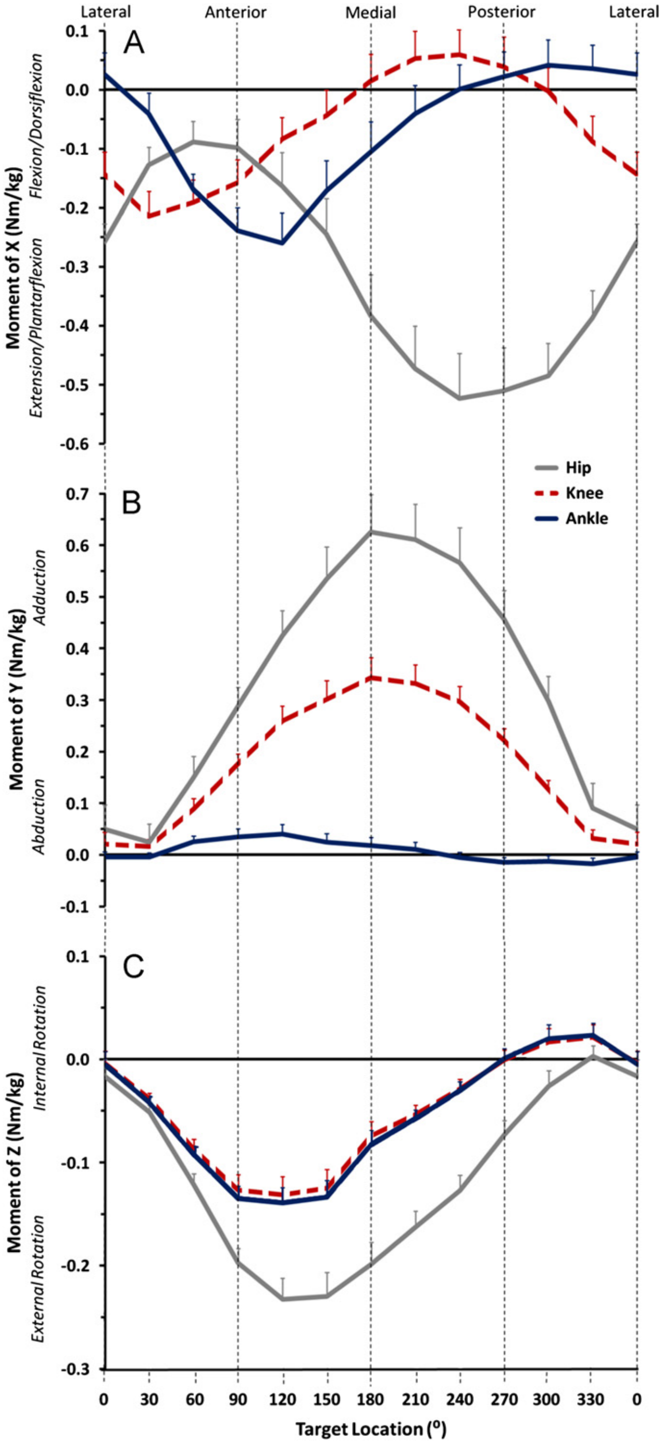

Fig. 2. Group mean and standard errors of knee, hip, and ankle joint moments along the x-, y- and z-axes, respectively, expressed in newton metres and normalised to BM (Nm/kg). Values are mean taken over the 1 s of target match.

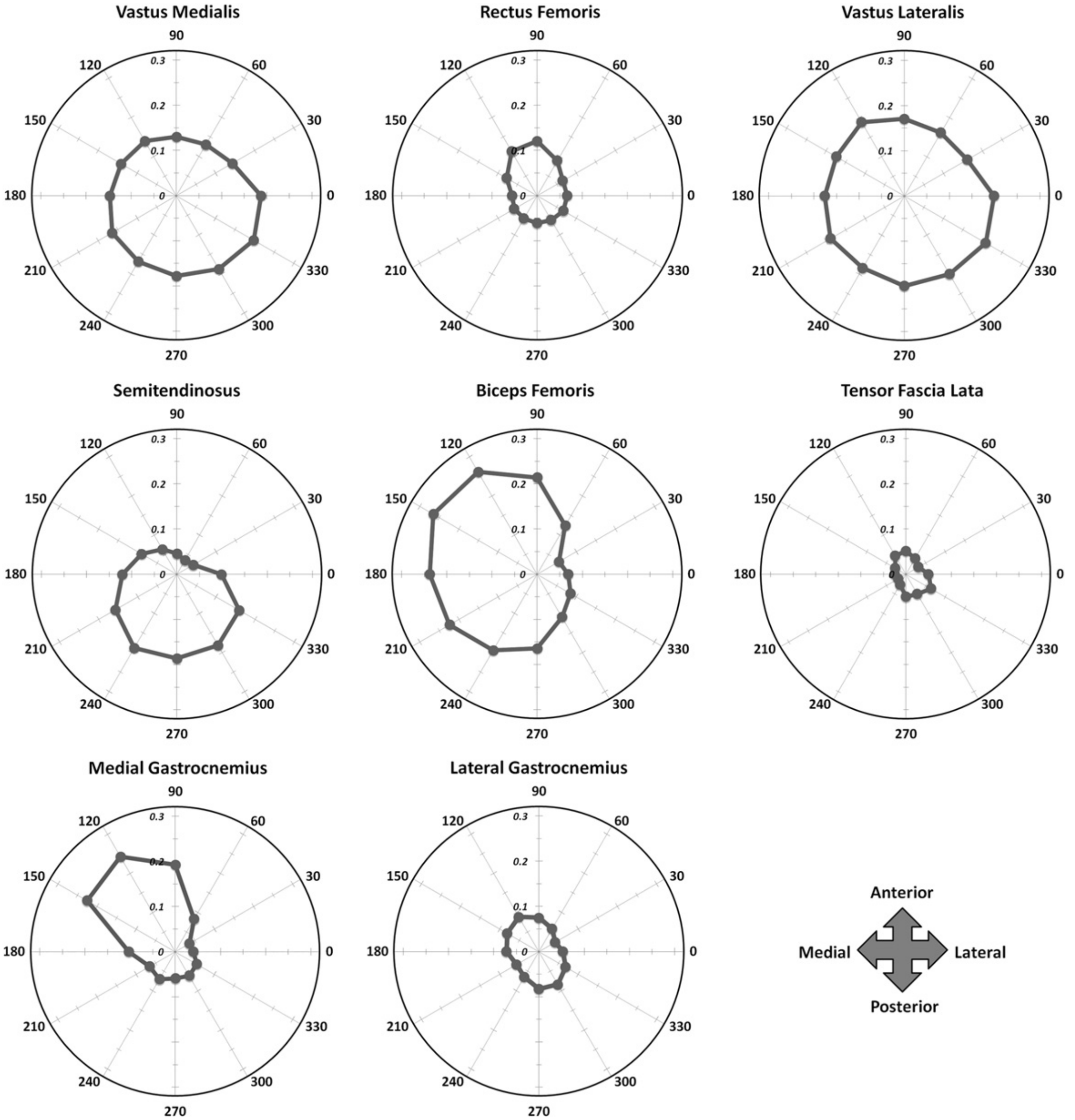

Fig. 3. EMG polar plots of mean activation patterns. Outer numbers along the circular trajectory represent the target location angle (°). Inner numbers along each radius represent normalised EMG magnitude. All plots are scaled from 0.0 to 0.30 representing 0–30% MVIC EMG magnitude. Where the pattern on the target location's radius intersects represents the mean normalised EMG utilised to match that target.

Table 1
The intra-class correlations ($ICC_{(2,\ k)}$) and confidence intervals (CI) of each muscle indicating the variability of subject activation profiles compared to the mean activation profile.

Muscle	RF	VL	VM	BF	ST	LG	MG	TFL
ICC	0.876	0.783	0.904	0.977	0.977	0.816	0.964	0.846
Upper CI	0.957	0.926	0.967	0.992	0.992	0.937	0.988	0.947
Lower CI	0.745	0.555	0.803	0.952	0.952	0.622	0.926	0.683

4. Discussion

This study explored voluntary knee muscle activation patterns of young healthy male adults while weight-bearing. Subjects isometrically produced GRF ranging from 0.48 to 0.58 (± 0.17–0.19) N/kg while maintaining 50% BM on the test limb. These GRF are comparable to those measured during ADLs such as the maximum anterior–posterior GRF in running initiation and termination (0.3 N/kg and 0.4 N/kg, respectively) and the medial-lateral GRF in jump landing (medial = 0.4–0.7, lateral = 0.1–0.4 N/kg) (McClay et al., 1994; Smith et al., in press). This illustrates the external validity of our task and we therefore believe physiologically relevant neuromuscular control strategies are being observed.

Non-weight-bearing isometric force matching studies indicate that muscle activation is directly related to its moment arm orientation, leading to asymmetrical activation patterns and high muscle specificities (Buchanan and Lloyd, 1997; Krishnan et al., 2008; Lloyd and Buchanan, 2001; Williams et al., 2003). However, we found low measures of specificity in all muscles (Fig. 4B), and symmetrical activation patterns in three muscles (VL, VM, and LG) (Fig. 3). Our activation patterns are consistent across subjects (ICCs > 0.75; Table 1) and cannot be attributed to randomness. Only ST demonstrated an activation profile similar to the previous muscle specificity research (Fig. 3). Additionally, we observed higher muscle activation levels (X_{EMG}; Fig. 4C) in the quadriceps and hamstrings compared to previous force matching studies. We attribute these differences to our standing weight-bearing protocol, as the addition of body weight alters feedback mechanisms,

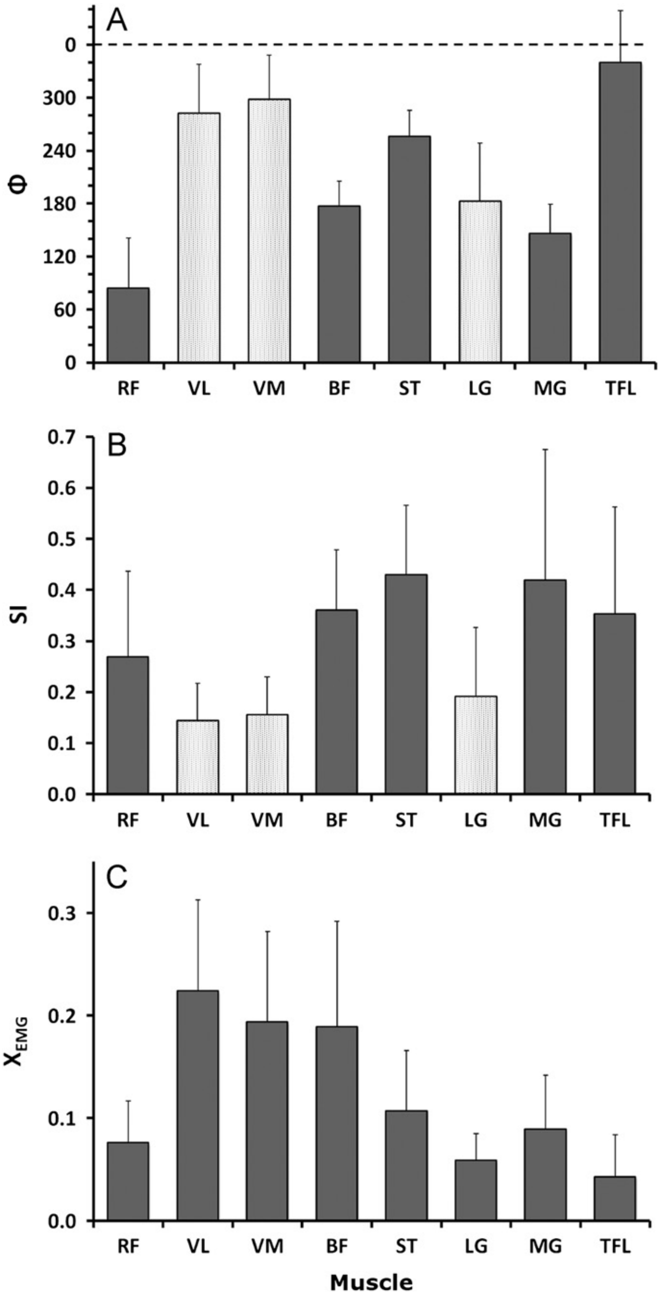

Fig. 4. (A) Mean direction of muscle activation (Φ) expressed in degrees (°), (B) specificity indices (SI), and (C) mean magnitude of muscle activation (X_{EMG}) of eight muscles crossing the knee joint. EMG values are normalised so they range from 0.0 to 1.0 with 1.0 being the maximal recorded EMG value for that muscle. Error bars indicate standard deviation. Note: The VL, VM, and LG demonstrated symmetrical activation patterns (statistically equal activation in all directions). As such, their Φ and SI bars are unfilled and should not be interpreted as meaningful.

lower limb kinematics, and muscular recruitment characteristics (Escamilla et al., 1998; Kiefer et al., 1998; Shultz et al., 2009; Wilk et al., 1996).

Based on rigid body mechanics, when forces applied across a joint are balanced adjoining segments will remain in a static position, whereas unbalanced forces will cause changes in rotation and/or translation at the joint. During our tests, subjects stood and maintained static joint positions, meaning the lower limb acted as a rigid segment interconnected at the knee. Considering dominant hip moment strategies were elicited for generating medial, posterior and lateral loads, we postulate that most knee muscles served as joint stabilisers, contracting to maintain static equilibrium so hip muscles could generate and transmit the moments needed to match the cursor to the target through the knee to the foot-ground interface. Without these stabilising contributions at the knee, moments acting on the proximal femur would significantly increase loads exposed to the knee's soft tissue restraints and increase the risk for traumatic injury (Fleming et al., 2001; Fujiya et al., 2011).

We hypothesised that muscle activation patterns would be dictated by knee joint stabilisation strategies rather than MAO, which is supported by our muscle activation patterns and hip and knee joint moments. As such, we propose three major roles for knee joint muscles during standing weight-bearing:

(1) General joint stabiliser: a muscle with a symmetrical activation pattern.
(2) Moment actuator: a muscle with relatively high specificity and an asymmetrical activation pattern about its reported MAO.
(3) Specific joint stabiliser: a muscle with relatively high specificity and an asymmetrical activation pattern opposite of its reported MAO.

Accordingly, VL and VM were classified as general joint stabilisers; each vastii muscle activated at similar magnitudes in all GRF directions. We believe these synergistic activation strategies increase joint compression and stability, rather than facilitate the modulation of GRF direction. Their symmetrical activation patterns suggest that they stabilise against hip adduction and hip extension loads that would, respectively, cause valgus alignment and posterior femoral translation. This observation is supported by Earl et al. (2001) and Hertel et al. (2004) who found that vastii activation levels did not differ when hip adduction or abduction components were added to a squatting exercise.

In contrast, RF demonstrated an anteriorly oriented asymmetrical activation pattern and was classified a moment actuator. While it displayed a relatively low X_{EMG} ($7 \pm 2\%$ MVIC), it has a large cross sectional area and may still represent functionally significant contributions to moment generation (Brand et al., 1986). In addition, RF activation may have been underestimated since we normalised it to a seated task. Unlike the vastii muscles, the bi-articular RF activation is dependent on hip position, which is greater during seated knee extensions compared to squatting or leg presses (Kong and van Haselen, 2010; Maffiuletti and Lepers, 2003). Considering this, we believe the role of bi-articular muscles in lower limb control is important to discuss (van Deursen et al., 1998; van Ingen Schenau et al., 1992). For example, subjects used hip extension to reach posteriorly located targets and minimal knee joint moments. Thus, what occurs at the knee depends on the relative contribution of the mono- and bi-articular hip extensors (e.g. gluteus maximus vs. hamstrings). If the hamstrings primarily activate, then relatively more knee extension moment is needed to counteract the hamstring's knee flexion moment. Conversely, if the gluteus maximus primarily activates, no hamstring knee flexion moment results and opposing knee extension is not required. As such, we speculate that the hamstrings primarily acted as hip extensors and their corresponding knee flexion moment was opposed by the vastii. Similar analyses can be made for one- vs. two-joint muscles in any other direction, as well as muscles acting on the knee and ankle. Of course, altering one's strategy to generate only knee moments is plausible, and then hip muscles become stabilisers, but this was not observed in our population.

The ST was also deemed a moment actuator because its relatively specific activation pattern occurred about its reported moment arm orientation MAO (Buchanan and Lloyd, 1997), similar to previous studies which investigated the movement facilitation capabilities of knee muscles (Buchanan and Lloyd, 1997; Krishnan et al., 2008; Lloyd and Buchanan, 2001; Williams et al., 2003).

In contrast, the BF's activation pattern was specific about a medial loading direction that is opposite of its reported MAO (Buchanan and Lloyd, 1997). Medial loads were mainly generated by hip adduction, which induces a valgus moment at the knee. However, muscles with substantial varus moment arms (e.g. sartorius and gracilis) (Buchanan and Lloyd, 1997) only contribute 5.5% to an opposing varus moment (Lloyd and Buchanan, 2001). In addition, hamstring-quadriceps cocontraction only influences anterior tibial shear forces and internal rotation during weight-bearing (MacWilliams et al., 1999). Therefore, BF is classified as a specific joint stabiliser that opposes hip extension, and more importantly, valgus loads. In support of this observation, deficits in lateral musculature have been correlated to greater valgus deviation and external rotation (Hanson et al., 2008; Shultz et al., 2009).

The LG was classified as a general joint stabiliser but demonstrated relatively low activation in all directions. The MG's activation pattern was specific about an anterior-medial loading direction that did not correspond with its reported moment arm orientation (Buchanan and Lloyd, 1997), classifying it as a specific stabiliser. It must be noted, however, that the MG's activation patterns corresponded to the plantar flexion moments during anterior-medial loads. As such, we exercise caution when speculating on the gastrocnemius' role as it relates to knee joint stabilisation, even though its activation alters in subjects with unstable knees during gait (Benoit et al., 2003) and may affect anterior tibial translation (Fleming et al., 2001). Distinguishing between the gastrocnemii's plantar flexor role and their influence on knee stabilisation requires further investigation, but their respective activation patterns, as well as those within the quadriceps and hamstrings, underline the importance of analysing muscles independently within their functional group.

Lastly, previous works indicate that the TFL contributes up to 25% of muscular loads to lateral knee compartment stabilisation during gait (Winby et al., 2009) and assist in knee extension and minor valgus moment generation (Buchanan and Lloyd, 1997; Lloyd and Buchanan, 2001). Our TFL's activation patterns, specific to lateral loading, had low X_{EMG} levels ($4 \pm 1\%$ MVIC) which, in our opinion, do not allow us to speculate on it role.

In addition to limitations addressed previously (Smith et al., in press), our interpretation of muscular contributions to lower limb force and moment generation was limited to eight muscles. Since research has shown that smaller muscles crossing the knee (vastus intermedius, semi-membranosus, short head biceps femoris' short head, sartorius, and gracilis) contribute to knee joint moments (Buchanan and Lloyd, 1997) and gluteal muscles, hip abductors and adductors are large determinants of femoral orientation (Geiser et al., 2010; Hanson et al., 2008), examining a more complete set of muscles that affect knee joint loading seems warranted. Our task was also kinematically isometric and thus results may not be directly applied to dynamic conditions. Finally, we did not specify physical activity type, experience level or frequencies in our subject exclusion criteria, even though knee muscle activation varies as a function of physical activity level (da Fonseca et al., 2006) and type (Lattier et al., 2003). This may explain our high between subject variation of X_{EMG} levels. However, we observed similar activation profiles across all subjects (indifferent of EMG magnitudes) indicating the muscle synergies were maintained throughout our subjects. We therefore can use our method to study different populations, since a deviation from the "healthy", reliable activation profile could provide insights into neuromuscular function associated with other populations.

5. Conclusions

Our study presents novel information on neuromuscular function as it relates to knee joint stability while weight-bearing. GRF generated are comparable to those of ADLs and we can identify the individual muscular contribution associated with each direction of force. It is evident from our study that knee joint neuromuscular control strategies are more complex than previously theorised and cannot be easily ascertained based on MAO. We therefore classified the roles of muscular support as: (1) general joint stabilisers (VL and VM), (2) specific joint stabiliser (BF), and (3) moment actuators (ST and RF). We suggest stabilisers increase joint compressive forces, creating a stable mechanical linkage at the knee from which moment actuators can initiate directed forces. This may be key in improving musculoskeletal models that attempt to estimate muscle activations since they would need to account for this stabilising role, as well as rehabilitation programs focused on knee joint stabilisation. Our results also demonstrate the importance of utilising weight-bearing conditions and analysing muscles independently within a muscle group.

Conflict of interest statement

This research was funded by the Natural Sciences and Engineering Research Council, the Canadian Foundation for Innovation, the Province of Ontario and the University of Ottawa Faculty of Health Sciences. To our knowledge, no conflicts of interest exist.

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