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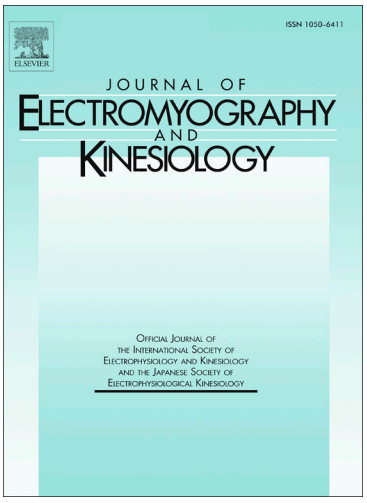

Representing fine-wire EMG with surface EMG in three thigh muscles during high knee flexion movements

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Title: Representing fine-wire EMG with surface EMG in three thigh muscles during high knee flexion movements

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Abstract:
Activation waveforms of vastus intermedius, adductor magnus, and semimembranosus have not been reported for high knee flexion activities such as kneeling or squatting, likely due to the invasive procedures required for their measurement. Their relatively large physiological cross sectional areas would suggest their contributions to knee joint loading could be considerable. Therefore, the purpose of this study was to quantify the activities of these muscles using fine-wire EMG and to assess easy to measure surface sites (vastus lateralis, rectus femoris, vastus medialis, semitendinosus, and biceps femoris) for their potential as proxy measures using < 10 %MVC RMS and > 0.85 R^2 as criteria for successful representation of deep muscle activity by that measured at a surface site. Overall, no surface and fine-wire site pair met both criteria for these movements. When fine-wire measurement of muscle activity is infeasible or impractical, the waveforms presented in supplementary material could be used as a guide for the activity of these deep muscles. Although select muscles for some participants satisfied our criteria, inter-participant variability was considerable. Therefore, future muscle models may benefit from fine-wire measurement of these muscles, but researchers should be cautious of electrode site specificity.

Keywords: kneeling, squatting, indwelling, fine-wire, electromyography

1. Introduction

Muscular activation linear envelopes of the vastus intermedius (VI), adductor magnus (AM), or semimembranosus (SM) are unknown during high knee flexion movements. High knee flexion is defined as movements where knee flexion exceeds 120º (Hemmerich et al., 2006; Kingston and Acker, 2018; Zelle et al., 2009). Activation waveforms for these muscles are needed for muscle force modeling in high knee flexion postures. Previous work, that was unable to measure deep musculature, represented the VI waveform as the average of vastus medialus (VM) and vastus lateralus (VL) or semitendinosus (ST) as identical to SM (Lloyd and Besier, 2003). Similarly, optimization based musculoskeletal models currently have limited (Byrne et al., 2005; Montgomery et al., 1994; Saito et al., 2015) or no (Hamner et al., 2010; Martelli et al., 2015) verification data to assess biofidelity of predicted muscle activity in these three large muscles.

There have been previous attempts to represent fine-wire activation waveforms from surface EMG data. Jacobson et al. (1995) measured VM and biceps femoris (BF) activity during walking and running from 12 males with both surface and fine-wire electrodes. Between the two sites, there were similar variance ratios (< 0.4), reproduceability, and linear envelope shapes (R^2 > 0.85) overall. McGill et al. (1996) reported that, in 5 males and 3 females, surface measured muscle activity could represent fine-wire measured activity of the quadratus lumborum, external oblique, internal oblique, and transverse abdominis muscles within 15% RMS but stated their R^2 comparisons were not informative as phase misalignment of EMG peaks can lead to

unexpectedly low values given good overall visual agreement. Byrne et al. (2005) found a modest linear correlation ($r = 0.579$, $R^2 = 0.336$) between surface and fine-wire recordings of the rectus femoris (RF) concluding that surface recordings might not be representative of RF activation levels due to vastii crosstalk. Finally, Allen et al. (2013) compared surface and fine-wire activity from supraspinatus and infraspinatus, in 10 males and females, during a number of isometric exertions. During external or internal axial humeral rotation trials respectively, surface recordings overestimated supraspinatus by 32% ($R^2 = .76$) and 21% ($R^2 = 0.72$) and infraspinatus by 72% ($R^2 = 0.64$) and 500% ($R^2 = 0.62$) (Allen et al., 2013). Although these previous studies have achieved varying levels of success in using surface recordings as proxies for fine-wire recordings, the strong R^2 findings of Jacobson et al. (1995) were the primary motivation for this study and were used to establish our R^2 criterion.

The purpose of this study was to quantify the activation of VI, AM, and SM using fine-wire electrodes and to compare these signals to those acquired from easily accessible surface locations over VL, RF, VM, ST, and BF. We hypothesised that relationships exist in which fine-wire signals may be estimated reliably from surface sites. Two criteria were used to evaluate if the surface locations reliably represented fine-wire: Coefficient of determination (R^2) greater than 0.85 (Jacobson et al., 1995) and RMS difference less than 10% MVC (McGill et al., 1996). These relationships, if robust, would simplify future work into muscular control in high knee flexion movements and could potentially improve musculoskeletal model estimates of knee joint contact forces.

2. Materials and Methods

2.1. Participants

Sixteen participants, eight male and female, were recruited as a sample of convenience from the university's study body (Table 1). Exclusion criteria consisted of any low back or lower limb injury within the past year that required medical intervention or time off from work for longer than three days, and any history of surgical interventions to the back or lower limb. All participants self-reported right leg dominance and the ability to kneel to the ground without pain. Each participant read and signed an informed consent form approved by the university's research ethics board.

2.2. Experimental procedures

Participant height and segmental anthropometrics were measured from the right lower limb before instrumentation (Table 1). Participant mass was calculated from force plate data during a static calibration trial. Thigh length was defined as the distance from the greater trochanter to the lateral femoral condyle. Distal from the greater trochanter, proximal circumference was measured at 10%, mid at 50%, and distal at 90% along this length. The participant's right leg was then instrumented with wireless surface EMG equipment (Wave Plus, Cometa srl, Milan, IT; input impedance = 20 MΩ, common mode rejection ratio = 120 dB at 60 Hz, bandpass filter 10-1000 Hz) to measure activity of the VL, RF, VM, ST, and BF at 2100 Hz. Electrode sites were located and prepared following SENIAM guidelines (Hermens et al., 2005) in a similar configuration to a previous high knee flexion study (Kingston et al., 2016). Bipolar Ag/AgCl electrodes (BlueSensor N, Ambu Inc., Glen Burnie, MD, USA) were adhered, with 2 cm inter-electrode spacing, after shaving, abrading, and cleaning of the skin. Surface electrodes with inter-electrode spacing between 2 and 2.5 cm were attached over the AM and SM insertion sites (described in the following paragraph). This spacing was somewhat variable from person to person to avoid interference with inserted fine-wires.

Following initial surface EMG preparations, fine-wire electrodes were inserted into the VI, AM, and SM (Figure 1) of the right leg and wirelessly recorded at 2100 Hz using the same equipment as surface signals. Researchers wore new nitrile gloves for each insertion and used isopropyl alcohol to create a 2 cm^2 sterile field at the insertion site. Sterile single-use 50 mm long 25 gauge (0.55 mm) hypodermic needles (Motion Lab Systems, Inc., Baton Rouge, LA) were used to insert bipolar fine-wire electrodes using guidelines from Perotto (2011) and real-time ultrasonography (M-Turbo, Sonosite Inc., WA, USA; Figure 1). Each needle contained two nylon insulated 304 series stainless steel wires (0.051 mm x 200 mm), which were insulated and had a 2 mm exposed sensor—with hooked ends—inside the muscle and 5 mm bare-wire terminations for connection to spring leads. Fine-wires extended > 8 cm beyond the surface of the skin (Figure 2). Following each fine-wire insertion, participants firmly contracted against manual resistance 3-6 times in knee flexion/extension (VI and SM) and hip adduction (AM) to set fine-wires inside the muscle, and were able to stand and walk if cramping or discomfort occurred, until they self-reported that discomfort had subsided. Fine-wires remained in muscles for approximately 1 (SM) and 1.75 (VI and AM) hours.

Participants sat on the edge of a massage table (~90º knee flexion) for VI and AM insertions. Fine-wires for VI passed through the rectus femoris (RF) and terminated at the mid-point of the muscle belly (Figure 1 A). Prior to the insertion of AM fine-wires Doppler ultrasound was used to identify femoral artery blood flow, then gentle adductions of the femur was monitored via ultrasound to identify the gracillis, AM, and adductor longus muscles (Figure 1 B).

Participants then completed two 6 s isometric maximum voluntary contractions (MVC) for each muscle group with a minimum 60 s rest between trials. Vastii MVCs were performed with the right leg in a commercial leg extension exercise machine, under isometric conditions, with

the knee joint positioned at 45º of flexion (Hermens et al., 2005; Kingston et al., 2016). Adductor MVCs were performed with participants seated on a massage table where they isometrically adducted their hips to squeeze the thorax (~0.5 m diameter) of the investigator.

Participants were prone for SM insertions. Prior to the insertion of SM the popliteal artery was identified with Doppler ultrasound medial to the semitendinosus tendon, then gentle knee flexion contractions were performed to find the border between SM and the flexor head of AM (Figure 1 C). Semimembranosus MVCs were performed isometrically against manual resistance with the knee at 65º of flexion (Hermens et al., 2005; Kingston et al., 2016).

After EMG preparations, rigid bodies were attached to the right thigh, shank, foot, and the pelvis for kinematic tracking (Figure 3). The following anatomical points were digitally reconstructed for each segment: For the thigh, greater trochanter, and medial/lateral femoral condyles; for the shank, medial/lateral tibial condyles, tibial tuberosity, and medial/lateral malleoli; for the foot, medial/lateral malleoli, $1^{st}/5^{th}$ distal metatarsal heads, and heel; and for the pelvis, left/right anterior superior illiac spine, left/right posterior superior illiac spine, left/right illiac crest, and sacrum. Kinematic data were recorded at 100 Hz using an optoelectronic system (Optotrak, NDI, Waterloo, ON). Kinetic data were recorded at 2100 Hz from two embedded force plates (OR6-7, AMTI, Watertown, MA). All data were synchronized via collection software (First Principles, NDI, Waterloo, ON) with a fixed 14 ms telemetric delay in EMG data accounted for in data processing.

Participants then completed a static standing trial, followed by knee and hip functional joint center trials (Besier et al., 2003; Camomilla et al., 2006). The high knee flexion movements in this study were the same as those used in a previous study (Figure 3) by Kingston and Acker (2018). Participants first observed all movements being performed by the investigator and then

practiced until they could perform each movement comfortably. One repetition of each movement and a single walking trial were completed in fixed order. Then order was fully randomized for four more repetitions (for a total of 5 repetitions in each movement) of: heels-up squat (HS), flatfoot squat (FS), dorsiflexed kneel (DK), plantarflexed kneel (PK), dorsiflexed unilateral kneel (DUK), plantarflexed unilateral kneel (PUK), and walking (WK). The fixed order block was used to ensure that at least a single trial of each movement was recorded as quickly as possible in case of accidental fine-wire shift or discomfort. Each squatting or kneeling trial took 6 s to complete and consisted of stepping onto an embedded force plate, descending to maximal knee flexion, and holding the position. Walking trials began with participants two steps away from the force plates such that their third step was contact of the right foot on a single force plate. Participants moved at a self-selected pace in all trials, with the following movement restrictions during high flexion movements: step with the right foot first; kneel onto the right knee (kneeling trials); then hold the final posture until instructed to stand. During performance of DUK or PUK, participants were instructed to shift the most of their bodyweight onto the right leg to resemble firing positions used in military theater (Department of the Army, 2010).

Data Processing

Processing was completed using Matlab 9.2 (R2017a, The Mathworks, Natick, MA). Kinematic and ground reaction force (GRF) data were low-pass filtered using a bidirectional 2nd-order Butterworth digital filter with a 6 Hz cut-off frequency (Longpré et al., 2013; Winter, 2009). Knee and hip joint centres were calculated from functional trials using the Symmetrical Center of Rotation Estimation (SCoRE) algorithm (Ehrig et al., 2007, 2006) which provides accurate hip joint centre predictions from skin markers when compared to dual-plane fluoroscopy (Fiorentino et al., 2016). Knee joint angles were decomposed in a flexion/extension-

ab/adduction-axial rotation Cardan sequence (Wu and Cavanagh, 1995). Data were then truncated from vertical GRF component exceeding 10 N to a manually identified frame where the knee flexion waveform plateaued in high flexion movements (Kingston and Acker, 2018) and from heel-strike to toe-off in walking. Activation waveforms were visually screened for motion and/or electrode contact artifacts, then processed using a 2 Hz low-pass single-pass Butterworth filter to produce a linear envelope and normalized to isometric MVCs (Kingston et al., 2016). The activation waveform of VI was compared to three surface vastii sites (VL, RF, VM), with SM compared to three surface hamstring sites (surface SM, BF, ST), and AM compared to its surface site. Time normalized trials were averaged within participant with RMS differences calculated between fire-wire and surface activation waveforms. RMS differences were then averaged across participants (Chapman et al., 2010; McGill et al., 1996). Regression was performed within participant between fine-wire and respective surface sites using a least-squares quadratic polynomial to define our R^2 criterion (Allen et al., 2013; Byrne et al., 2005; McGill et al., 1996) and then averaged across participants.

3. Results

Based on mean RMS differences and R^2 values, no surface sites satisfied either of our criteria (< 10% MVC RMS or $R^2 > 0.85$) to act as a proxy for fine-wire sites in these movements. Mean RMS and R^2 of our sample population in each movement are reported in Table 2. The best matched surface and indwelling signals from our sample, as per our stated criteria, are shown in Figure 4. For reporting purposes only, please see Supplementary Material (A1-7) for across participant mean fine-wire activation profiles, normalized to knee flexion angle, for each movement.

4. Discussion

The purpose of this study was to quantify the activation of VI, AM, and SM using fine-wire electrodes for comparison to easily accessible surface sites. These comparisons took place for six high knee flexion activities and level walking using criteria of < 10% MVC RMS difference and >0.85 R^2 to indicate a sucessful surface to fine-wire relationship. None of the surface sites satisfied our criteria across this healthy young sample. This is largely due to the considerable variability of surface-indwelling comparisons between participants (Table 2 and Supplementary Material). Our findings would suggest that the measurement of VI, AM, and SM muscles during high knee flexion movements cannot be accurately represented by surface sites and that the use of fine-wire EMG to obtain representative activation waveforms may be required if isolated muscle/motor unit activity is required.

Inherent to our research question was the effect of motion artifact in our surface EMG recordings compared to fine-wire signals. This issue is also listed within our limitations below. Our results would suggest that surface measurement of VI, AM, or SM is not robust to the effects of motion artifacts. Due to participants having performed movements using their entire range of knee flexion, the signal measured from surface EMG was not from the same volume of muscle fibers throughout the trial. Therefore, movement of muscle fibres could change the motor units, and respective EMGs, recorded. Surface EMG pickup volume would also be influenced by soft-tissue artifact as local displacement of electrodes is unavoidable in high knee flexion postures due to deformation of the thigh lean and soft tissues. We speculate that fine-wire measurements would be minimally influenced by motion or soft-tissue artifacts which could worsen the relationships measured to surface sites using our success criteria.

Although vastii musculature did not meet our criteria across this small sample on average, some individual participants met both criteria in select movements (primarily squatting

activities). At a sample level, results would suggest that VM is likely the only muscle that could be modeled if a more relaxed RMS and R^2 criterion could be accepted. For 5 of our 16 participants, VM satisfied our RMS criterion across all high flexion movements with 2 also satisfying our R^2 criterion in select cases. Interestingly, of the surface vastii comparisons, RF activation was the least representative of VI activation even though its line of action, and assumed mechanical function, is the most similar to VI.

The surface site for AM, confirmed appropriate via ultrasound, was below 20 %MVC for most participants in these activities while the fine-wire site was ~50 %MVC. We were surprised that the surface signal was lower than the indwelling given the influence of crosstalk from neighboring muscles due to the considerably larger pick-up volume of surface EMG compared to fine-wire (Basmajian and De Luca, 1985; Clancy et al., 2002; Winter et al., 1994). Even so, the AM comparisons were consistently the worst of the three fine-wire sites assessed in this study.

The RMS results of SM comparisons should be viewed with caution as the descent phase of high flexion activities generally requires less than 20 %MVC from hamstring muscles (Kingston et al., 2016); the small magnitude of the signals could allow this criterion to be met despite a poor fit in terms of pattern. Therefore, R^2 outcomes may be the more meaningful metric for this muscle group in this study. Across this sample, these muscles did not meet our R^2 criterion nor the more relaxed R^2 criterion (0.5) used by Jacobson et al. (1995).

The largest difference between surface and fine-wire sites always occurred during the weight bearing phase of our walking trial. We speculate that this is due to the localized pick-up volume of our fine-wire sites as the motor units with exposed sensors present may have been, by chance, far more active than the holistic representation surface sites provide (Clancy et al., 2002; Winter et al., 1994). The low physical demand of walking, in comparison to squatting or kneeling

performed in this study, may support this theory as site agreement would likely be higher if more motor units were recruited (Fuglevand et al., 1992; Henneman et al., 1965; Yao et al., 2000).

Limitations of this study include muscle fibre/motor unit movement relative to surface locations, the muscle fibre/motor unit type that was measured from fine-wire electrodes, and the relative novelty of some of these high knee flexion movements to most participants. As mentioned previously, the signal measured from surface EMG was not from the same volume of muscle fibers throughout the trial but we speculate that fine-wire measurement volumes would be minimally shifted. While fine-wire EMG provides a precise representation of muscle activity, we are not aware of any assessment (or the practicality) of the day-to-day repeatability in these measures for the muscles investigated. Finally, this sample of convenience consisted of young healthy individuals who do not commonly perform these high knee flexion movements. Therefore, the applicability of these findings to habitually kneeling populations (e.g. construction workers, East Asians, practicing Muslims) requires further investigation.

5. Conclusion

The results of this study suggest that between participant variability in %MVC RMS and R^2 is high when comparing surface EMG activation waveforms to fine-wire measurement of VI, AM, and SM during high knee flexion activities and walking. Therefore, representative surface locations were not identified for the high knee flexion movements investigated in this study. Future modelling efforts using Hill-type muscle force estimation may benefit from fine-wire measurement of the activity of these muscles, as crosstalk would be eliminated, but researchers should be cautious of electrode site specificity being unrepresentative of a musculotendinous unit.

6. Acknowledgements

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Captions to illustrations

Table 1: Mean (standard deviation) descriptive and anthropometric participant information. Circumferences were measured distally from the greater trochanter towards the lateral femoral condyle: proximal circumference was measured at 10%, mid at 50%, and distal at 90% of thigh length.

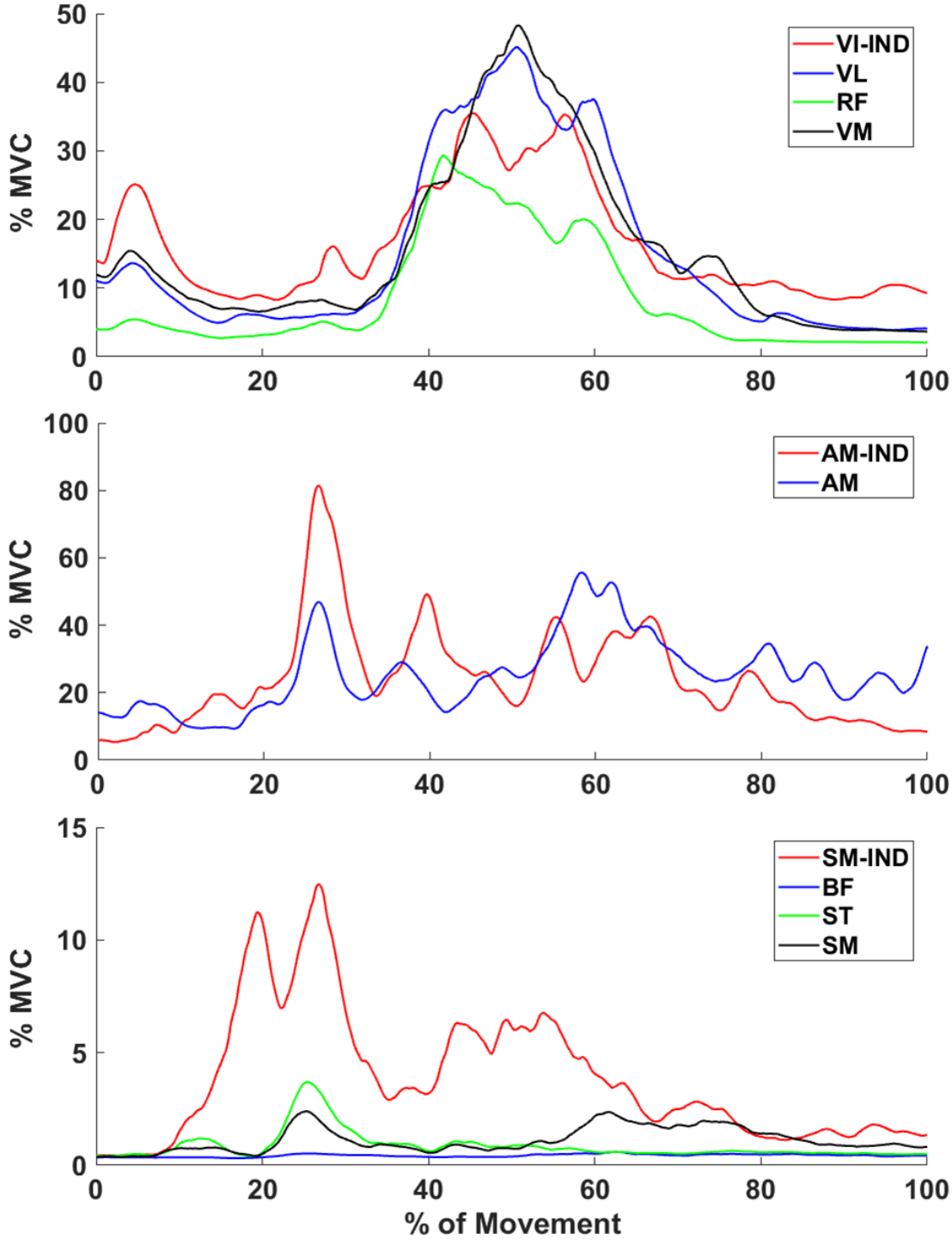
Table 2: Mean (standard deviation) of RMS and R^2 values across participants for high flexion movements for surface compared to respective fine-wire signals. Movements listed in the leftmost column are: heels-up squat (HS), flatfoot squat (FS), dorsiflexed kneel (DK), plantarflexed kneel (PK), dorsiflexed unilateral kneel (DUK), plantarflexed unilateral kneel (PUK), and walking (WK).

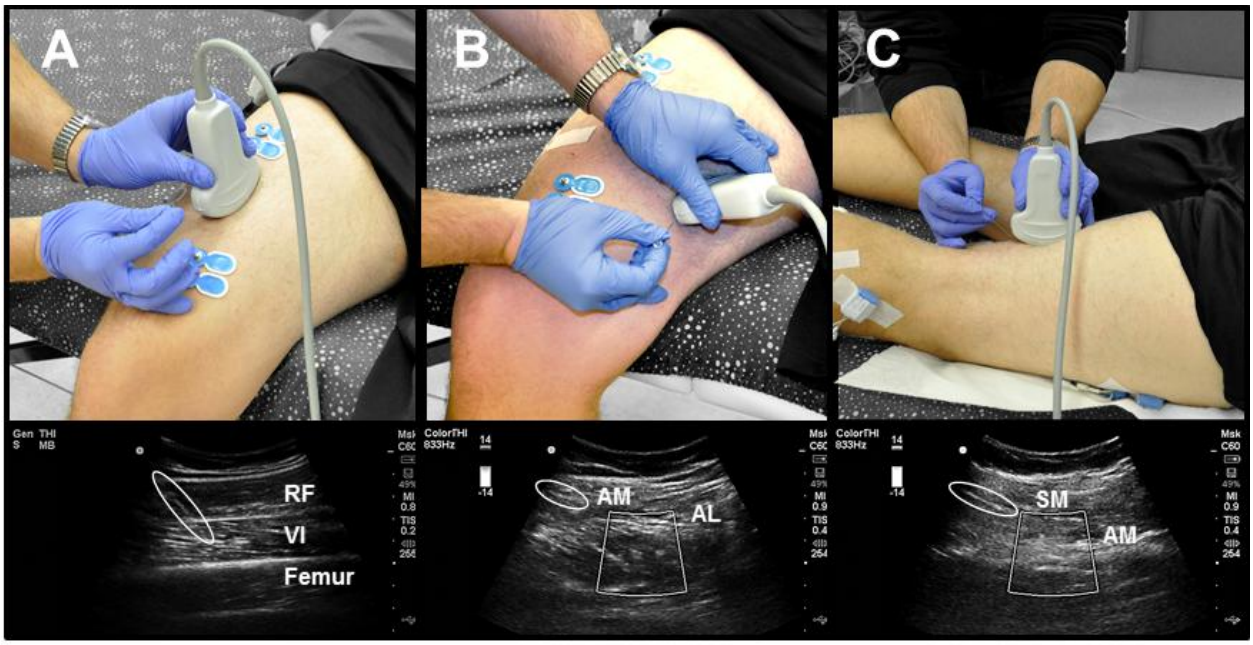
Figure 1: Fine-wire insertion locations and needle positioning during preparation of participant P16. Top row: Ultrasound probe placement and needle positioning for insertion. Bottom row: Needle location (circled) within muscles before the cannula was removed. RF is rectus femoris, VI is vastus intermedius, AM is adductor magnus, AL is adductor longus, and SM is semimembranosus.

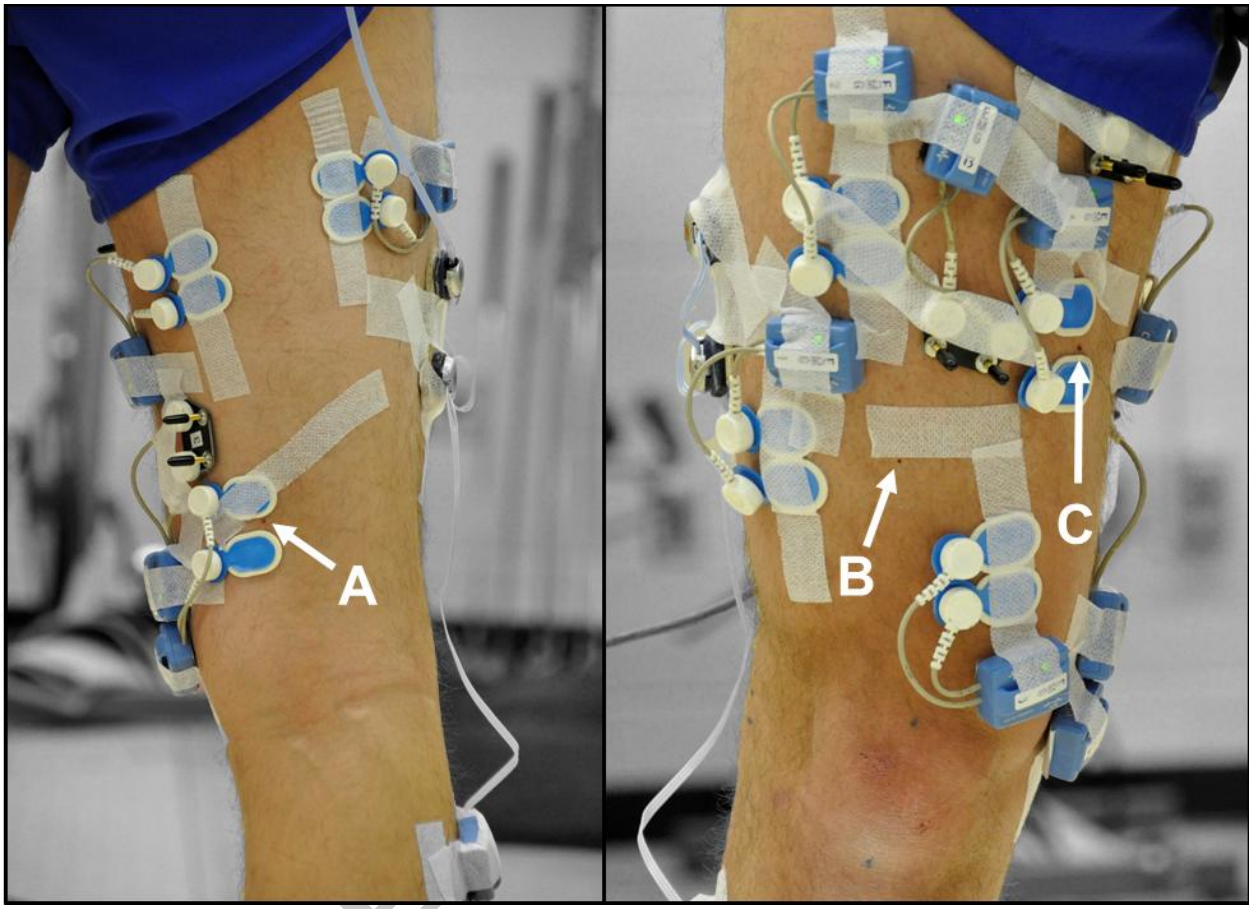
Figure 2: Fine-wire and surface EMG instrumentation from the posterior (left) and anterior (right) thigh of participant P04. Arrows indicate fine wire insertion sites. A) Fine-wire location of semimembranosus (SM) with surface EMG spanning the insertion site. B) Fine-wire location of vastus intermedius (VI). C) Fine-wire location of adductor magnus (AM) with surface EMG spanning the insertion site.

Figure 3: High knee flexion postures used in this study: (HS) Heels-up squat, (FS) flatfoot squat, (DK) dorsiflexed kneel, (PK) plantarflexed kneel, (DUK) dorsiflexed unilateral kneel, and (PUK) plantarflexed unilateral kneel. This figure has been modified from Kingston and Acker, (2018).

Figure 4: Muscle activation waveforms normalized to percentage of movement across five repetitions. Top: Vastii waveforms from participant P01 performing a dorsiflexed kneel. Middle: Adductor waveforms from participant P05 performing a heels-up squat Bottom: Hamstrings from participant P04 performing a flat-foot squat. Abbreviations used to indicate muscle sites are: fine-wire vastus intermedius (VI-IND), vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), fine-wire adductor magnus (AM-IND), adductor magnus (AM), fine-wire semimembranosus (SM-IND), semimembranosus (SM), semitendinosus (ST), and biceps femoris (BF).







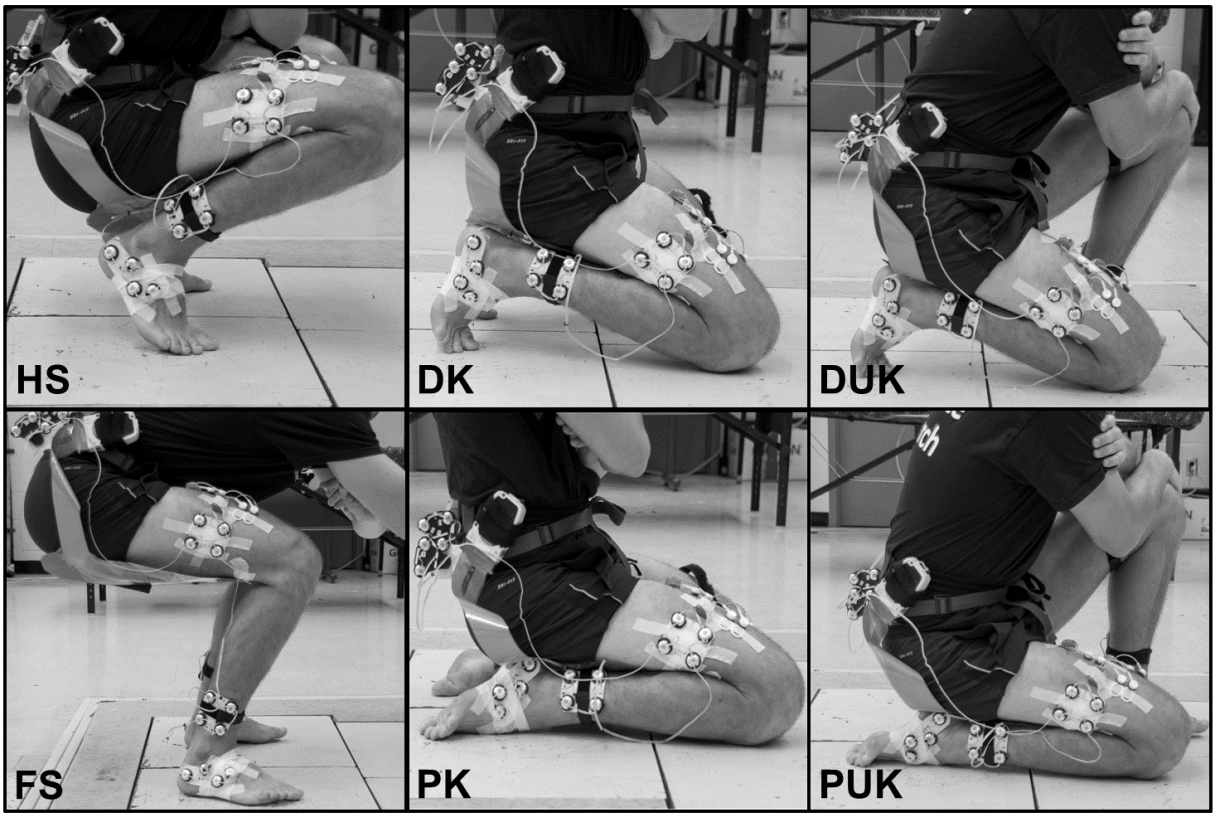


Table 1

Parameter	Females (*n* = 8)	Males (*n* = 8)	All (*n* = 16)
Age (yrs)	24.30 (4.5)	26.30 (3.2)	25.30 (3.9)
Height (m)	1.70 (0.1)	1.80 (0.1)	1.80 (0.1)
Mass (kg)	70.40 (10.7)	88.60 (16.5)	79.50 (16.4)
BMI (kg/m^2)	24.30 (3.8)	27.00 (3.4)	26.70 (3.8)
Thigh Length (m)	0.41 (0.04)	0.40 (0.04)	0.40 (0.04)
Thigh Proximal Circumference (m)	0.60 (0.04)	0.63 (0.09)	0.61 (0.07)
Thigh Mid Circumference (m)	0.53 (0.04)	0.55 (0.12)	0.54 (0.09)
Thigh Distal Circumference (m)	0.41 (0.04)	0.42 (0.05)	0.41 (0.04)

Table 2

Indwelling Surface	VI						AM		SM					
	Vl		RF		VM		AM		BF		ST		SM	
Moveme	RM	R^2	RM	R^2	RM	R^2	RM	R^2	RM	R^2	RM	R^2	RM	R^2
HS	15.6 (14.8)	0.40 (0.24)	16.3 (15.0)	0.29 (0.25)	14.7 (14.6)	0.39 (0.22)	40.7 (20.7)	0.39 (0.19)	27.3 (26.4)	0.29 (0.18)	26.6 (26.8)	0.36 (0.19)	26.9 (27.9)	0.31 (0.19)
FS	19.7 (17.3)	0.36 (0.33)	25.3 (23.5)	0.37 (0.34)	18.5 (16.6)	0.37 (0.35)	49.0 (23.0)	0.29 (0.20)	29.2 (29.3)	0.22 (0.21)	26.0 (26.9)	0.35 (0.2)	40.3 (63.9)	0.30 (0.22)
DK	17.2 (13.5)	0.46 (0.26)	17.7 (14.2)	0.45 (0.32)	15.4 (13.6)	0.45 (0.26)	43.0 (25.3)	0.32 (0.20)	21.6 (19.6)	0.18 (0.15)	20.8 (18.7)	0.30 (0.21)	35.5 (63.4)	0.39 (0.19)
PK	20.6 (20.6)	0.38 (0.22)	19.6 (20.2)	0.37 (0.27)	17.4 (19.7)	0.34 (0.17)	45.7 (24.6)	0.32 (0.16)	21.8 (17.8)	0.18 (0.14)	22.5 (19.9)	0.31 (0.19)	37.7 (63.8)	0.35 (0.22)
DUK	21.2 (19.4)	0.45 (0.26)	20.1 (18.7)	0.40 (0.29)	18.0 (18.9)	0.42 (0.25)	41.9 (22.5)	0.32 (0.15)	23.2 (22.2)	0.30 (0.20)	18.9 (12.5)	0.31 (0.22)	18.6 (12.2)	0.45 (0.19)
PUK	18.6 (15.8)	0.40 (0.32)	18.7 (15.6)	0.43 (0.33)	17.0 (15.1)	0.41 (0.30)	49.6 (34.3)	0.24 (0.16)	23.7 (26.8)	0.23 (0.15)	16.7 (13.0)	0.25 (0.25)	19.7 (19.1)	0.33 (0.24)
WK	39.3 (37.9)	0.43 (0.31)	36.3 (39.0)	0.34 (0.29)	35.2 (39.8)	0.43 (0.29)	76.1 (48.0)	0.45 (0.27)	33.5 (21.0)	0.38 (0.27)	32.8 (20.2)	0.52 (0.26)	34.7 (21.3)	0.48 (0.29)

Author Biographies

David C. Kingston is a PhD candidate in the Biomechanics of Human Mobility (BOHM) Laboratory at the University of Waterloo in Waterloo, ON, Canada. His PhD thesis is investigating the effect of intersegmental contact and EMG on knee joint loads during high flexion postures using a musculoskeletal model. He received his MSc and BSc in Kinesiology from Queen's University, in 2013 and 2011 respectively, with his undergraduate thesis winning a Sport Information Research Centre (SIRC) award in the High Performance category.

Stacey M. Acker is an Assistant Professor in Kinesiology at the University of Waterloo and directs research in the Biomechanics of Human Mobility (BOHM) Laboratory. She received her PhD in Mechanical Engineering from the Faculty of Applied Science at Queen's University, Kingston, Canada. She completed a postdoctoral fellowship in the MacMobilize Laboratory at McMaster University, Hamilton, Canada. Her research interests include lower limb joint modeling, occupational and orthopedic biomechanics, and osteoarthritis development and prevention. She is also a Researcher with the Centre for Research Expertise for the Prevention of Musculoskeletal Disorders.