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Lower-limb muscle function during sidestep cutting

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ABSTRACT

To investigate lower-limb muscle function during sidestep cutting, prior studies have analysed electromyography (EMG) data together with three dimensional motion analysis. Such an approach does not directly quantify the biomechanical role of individual lower-limb muscles during a sidestep cut. This study recorded three dimensional motion analysis, ground reaction force (GRF) and EMG data for eight healthy males executing an unanticipated sidestep cut. Using a musculoskeletal modelling approach, muscle function was determined by computing the muscle contributions to the GRFs and lower-limb joint moments. We found that bodyweight support (vertical GRF) was primarily provided by the vasti, gluteus maximus, soleus and gastrocnemius. These same muscles, along with the hamstrings, were also primarily responsible for modulating braking and propulsion (anteroposterior GRF). The vasti, gluteus maximus and gluteus medius were the key muscles for accelerating the centre-of-mass towards the desired cutting direction by generating a medially-directed GRF. Our findings have implications for designing retraining programs to improve sidestep cutting technique.

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1. Introduction

Sidestep cutting is frequently performed in sports such as football (Bloomfield et al., 2007), rugby (Green et al., 2011), and handball (Karcher and Buchheit, 2014). The ability to cut quickly and effectively is critical to overall performance in these sports. Sidestep cutting technique has also been linked with musculoskeletal injury, such as anterior cruciate ligament (ACL) injuries (Boden et al., 2000; Koga et al., 2010; Olsen et al., 2004). Less attention, however, has been placed on understanding the fundamental roles of individual lower-limb muscles during this type of change-in-direction manoeuvre. Such knowledge could be important for designing retraining programs to improve sidestep cutting technique.

The execution of locomotion tasks requires the coordination of multiple muscles, since no single muscle can perform all biomechanical functions (Zajac et al., 2002). Neptune and colleagues (Neptune et al., 1999) used surface electromyography (EMG) to provide insight into the role of multiple individual lower-limb muscles during sidestep cutting. However, inferring the biomechanical role of each muscle from EMG data alone is difficult, since

the way individual muscles contribute to joint and segment accelerations can sometimes be counter intuitive due to “dynamic coupling” (Zajac and Gordon, 1989).

Muscle actuated simulations have been used previously to predict muscle function during a variety of locomotion tasks (Dorn et al., 2012b; Hamner and Delp, 2013; Liu et al., 2008; Neptune et al., 2001). The contributions of individual lower-limb muscles to ground reaction forces (GRFs) or centre-of-mass accelerations can be used to understand how muscles achieve key biomechanical functions, such as bodyweight support and forward progression. Such analyses have mostly been limited to walking (Anderson and Pandey, 2003; Lim et al., 2013; Liu et al., 2006; Liu et al., 2008; Neptune et al., 2001; Pandey et al., 2010) and running (Dorn et al., 2012b; Hamner and Delp, 2013) in a straight direction. It is possible that lower-limb muscle function during sidestep cutting is distinct from that during walking and running, given the greater demands on accelerating the body's centre-of-mass in a medial direction. Subsequently, the aim of this study was to investigate lower-limb muscle function during a rapid sidestep cut. Specifically, we used a computational musculoskeletal modelling approach involving a GRF decomposition analysis (Dorn et al., 2012a, b; Lin et al., 2011) to determine muscular contributions to bodyweight support, forward progression and acceleration of the centre-of-mass in the desired direction of travel during an unanticipated sidestep cut.

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2. Methods

2.1. Participants

Eight recreationally healthy males (age, 27 ± 4 years; height, 1.77 ± 0.09 m; mass, 77.6 ± 12.8 kg) volunteered to participate in this study. All participants had no current or previous musculoskeletal injury likely to influence their ability to perform the required tasks. All participants provided written informed consent to participate in the study. Ethical approval was granted by the Australian Catholic University Human Research Ethics Committee (approval number: 2015-11H).

2.2. Instrumentation and procedures

Full details regarding the instrumentation and procedures have been reported previously (Maniar et al., 2018). Briefly, surface EMG data (1000 Hz) from 10 lower-limb muscles were synchronously recorded with full-body marker trajectories (200 Hz) and ground reaction force data (1000 Hz) whilst participants performed two alternative unanticipated cutting tasks. All participants completed the cutting tasks while barefoot to allow exposure of the foot for marker placement and to avoid any variability in the foot-ground interface. Participants were required to perform two single leg hops for a standardised distance of 1.35 m on their dominant (right) leg, and then as quickly as possible cut to the left (45-degree sidestep cut) or to the right (45-degree crossover cut) upon landing from the second hop. The direction of travel was randomly dictated by a set of timing gates that delivered a light signal ~ 450 ms prior to initial contact on the force plates. A successful trial required that the participant completed the task correctly with the entire foot landing within the force plate. Note that the 45-degree sidestep cut was the task of interest for this investigation, whereas the crossover cut was only included to ensure cutting direction was unanticipated.

2.3. Data processing

Marker trajectories were low-pass filtered using a zero-lag, 4th order Butterworth filter with a cut-off frequency of 8 Hz. This cut-off frequency was determined via a residual analysis. GRFs were filtered using the same filter and cut-off frequency as the marker data based on published recommendations (Kristianslund et al., 2012). EMG data were corrected for offset, high-pass filtered (20 Hz), full-wave rectified and low-pass filtered (6 Hz) using a zero-lag, 4th order Butterworth filter to obtain a linear envelope. EMG data were normalised to the peak amplitude obtained in each trial.

2.4. Musculoskeletal modelling

A 29 degree-of-freedom (DOF) full-body musculoskeletal model, with 80 musculotendon actuators (lower body) and 17 torque actuators (upper body) (Rajagopal et al., 2016), was used to perform the musculoskeletal simulations in OpenSim (Delp et al., 2007). Each hip was modelled as a 3-DOF ball-and-socket joint. Each knee was modelled as a 1-DOF hinge joint, with other rotational (valgus/varus and internal/external rotation) and translational (anteroposterior and superior-inferior) movements constrained to change as a function of the knee flexion angle (Walker et al., 1988). A pin joint was used to represent the ankle (talocrural) joint. The head-trunk segment was modelled as a single rigid segment, articulating with the pelvis via a 3-DOF ball-and-socket back joint. Each upper limb was characterised by a 3-DOF ball-and-socket shoulder joint and single-DOF elbow and radioulnar joints.

The generic model was scaled to each participant's individual anthropometry as determined during a static trial. An inverse kinematics algorithm was used to calculate joint angles by means of a weighted least-squares optimisation that minimised the difference between model and experimental marker positions (Lu and O'Connor, 1999). A residual reduction algorithm (RRA) was then used to make small adjustments to kinematics and torso mass properties to improve dynamic consistency between kinematic data and measured GRFs. As RRA requires a large number of user selected tracking weights for each DOF, these tracking weights were optimised using a particle swarm optimisation, to minimise joint angle tracking errors whilst also lowering the residual forces and moments at the pelvis (Samaan et al., 2016; Weinhandl et al., 2013). Muscle forces were obtained via static optimisation, which decomposed the RRA-derived joint moments into individual muscle forces by minimising the sum of muscle activations squared, taking into account the physiological force-length-velocity properties (Millard et al., 2013) of the musculotendinous units. This method of muscle force estimation is computationally efficient and has been used to predict muscle forces in similar high impact movements (Dorn et al., 2012b; Mokhtarzadeh et al., 2014, 2013). We then performed a GRF decomposition analysis (Dorn et al., 2012a, b; Lin et al., 2011) to determine muscular contributions to the GRFs.

2.5. Outcome variables

Muscular contributions to “support” as well as “braking and propulsion” are typically defined by their contributions to the vertical and anteroposterior GRFs, respectively. However, these definitions have been applied for planar tasks such as walking and running (Dorn et al., 2012b; Hamner and Delp, 2013; Liu et al., 2008; Neptune et al., 2001). Change of direction manoeuvres require appreciable acceleration of the body's centre-of-mass out of the sagittal plane, thus muscular contributions to the mediolateral GRF were also considered. Muscular contributions to GRFs were expressed in the global reference frame. Finally, consistent with other studies investigating muscle function (e.g. (Dorn et al., 2012b; Pandey et al., 2010)), we calculated muscular contributions to lower-limb joint moments.

Certain muscles were combined into functional groups consistent with prior research (Sriharan et al., 2012). Note that we only report on major muscle groups, and did not report on muscles that were not found to make meaningful contributions to the outcome variables (see (Rajagopal et al., 2016) for all musculotendinous actuators included in the model). We also note that swing limb muscles made no meaningful contributions to the GRF, thus we only report on stance limb muscles.

2.6. Validation and verification

Validation and verification of model predictions were performed in accordance with current best practice guidelines (Hicks et al., 2015). Qualitative comparisons between model-based predicted muscle activations and experimental EMG recordings showed good agreement after accounting for appropriate physiological delays of ~ 10 – 100 ms (Fig. 1). Additionally, the time-varying characteristics of our RRA-derived joint angles (Supplementary Fig. S1) and joint moments (Supplementary Fig. S2) were within 2SD of published data (Benoit et al., 2006; Oliveira et al., 2013; Sigward and Powers, 2006). Comparisons between experimental and simulated variables were evaluated via the normalised root-mean-square error (nRMSE) and coefficient of determination (R^2). The nRMSE was calculated as:

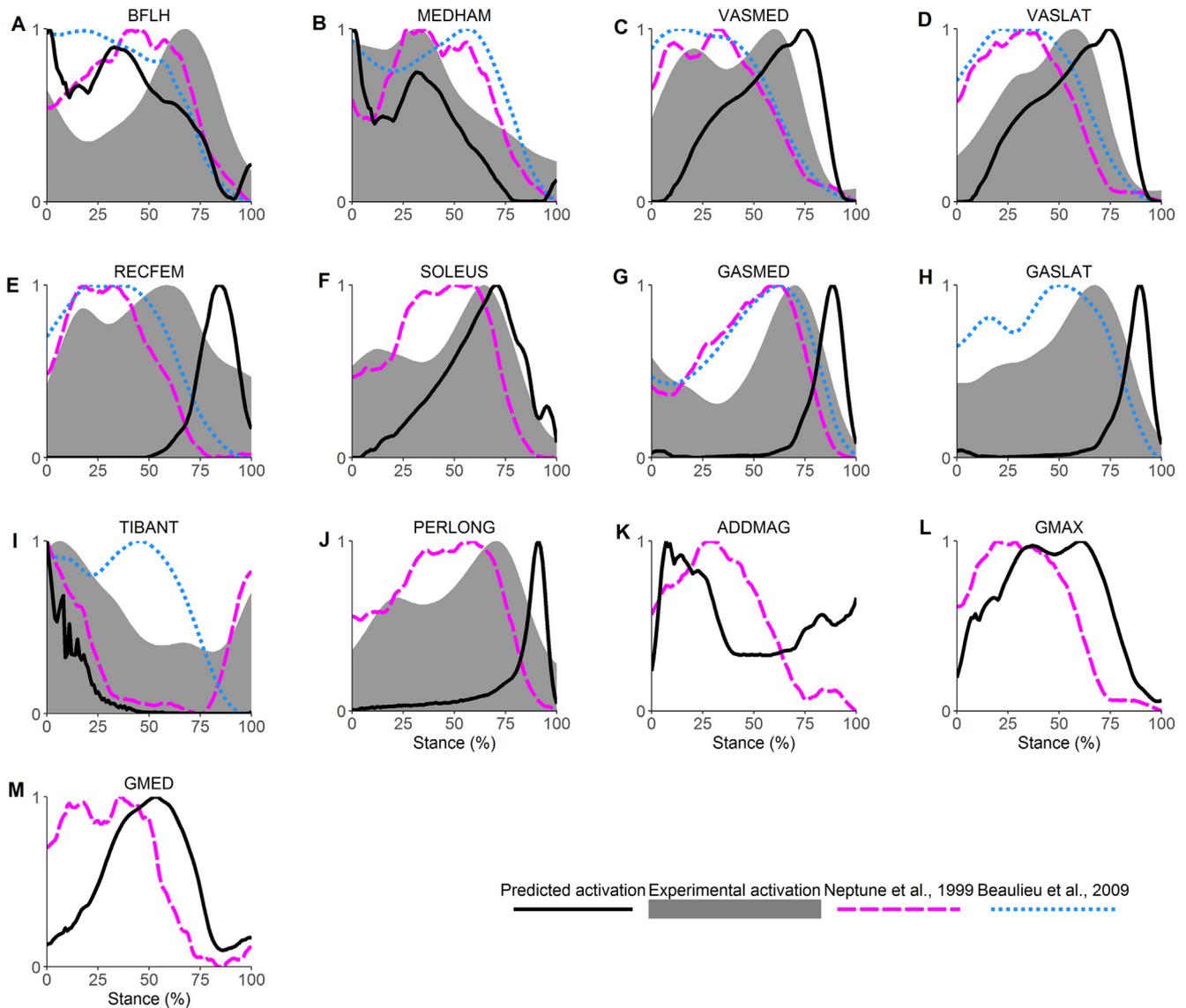


Fig. 1. Comparison of predicted (black line) and experimental activations (grey shaded) from the current data for the stance phase of the 45-degree unanticipated sidestep cut. Literature reference activations, magenta dashed line (Neptune et al. 1999); blue dashed line (Beaulieu et al. 2009). Panels, A, biceps femoris long head; B, medial hamstrings (semimembranosus and semitendinosus); C, vastus medialis; D, vastus lateralis; E, rectus femoris; F, soleus; G, gastrocnemius medialis; H, gastrocnemius lateralis; I, tibialis anterior; J, peroneus longus; K, adductor magnus; L, gluteus maximus; M, gluteus medius.

$$\text{nRMSE} = 100 \times \frac{\sqrt{\frac{\sum_{i=1}^n (\text{Experimental}_i - \text{Predicted}_i)^2}{n}}}{\max(\text{Experimental}) - \min(\text{Experimental})} \quad (1)$$

Muscle-derived joint moments (computed from the predicted muscle forces and their respective moment arms) were well matched with the experimental joint moments (median \pm interquartile range, $R^2 = 1.0 \pm 0.0$; $\text{nRMSE} = 2.0 \times 10^{-2} \pm 0.03\%$). Superposition errors between experimental and simulated GRFs were also well matched ($R^2 = 0.93 \pm 0.06$; $\text{nRMSE} = 9.8 \pm 3.7\%$). Residual forces and moments (Supplementary Fig. S3) and kinematic tracking errors (Supplementary Table 1) were also within recommended thresholds (Hicks et al., 2015).

3. Results

3.1. Braking and propulsion

Anteroposterior GRFs were characterised by a braking force (posteriorly-directed GRF) in the first half of stance, and propulsion

(anteriorly-directed GRF) in the second half (Fig. 2A). The antero-posterior GRF was primarily modulated by the hip extensors, knee extensors and the ankle plantar flexors. Specifically, braking throughout stance was primarily generated by the vasti (up to 335 N) and soleus (up to 151 N) muscle groups. The hip extensors were the primary contributors to propulsion for the first $\sim 60\%$ of stance, with the gluteus maximus and hamstrings producing up to 142 N and 102 N of the anteriorly-directed GRF, respectively. The contributions of these two muscle groups declined thereafter, with the ankle plantar-flexors taking over as the dominant contributors to propulsion (gastrocnemius, up to 312 N; soleus, up to 93 N).

3.2. Vertical support

Vertical support was primarily generated by the gluteus maximus, vasti, soleus and gastrocnemius (Fig. 2B). The gluteus maximus was the dominant contributor to vertical support in early stance, and produced up to 356 N of vertical GRF at $\sim 25\%$ of stance. This contribution declined thereafter. The vasti and soleus became

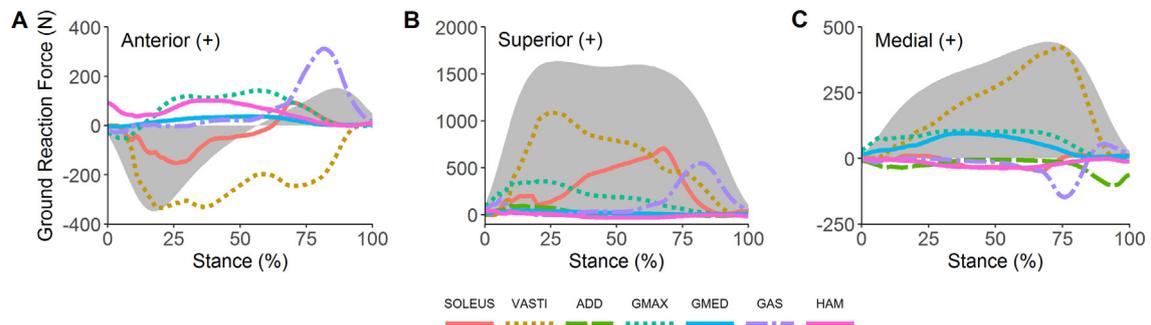


Fig. 2. Muscular contributions to ground reaction forces for the stance phase of the 45-degree unanticipated sidestep cut. Panel A, anteroposterior; B, superior-inferior; C, mediolateral. Note that the shaded grey represents the experimentally measured total ground reaction force. SOLEUS, soleus; VASTI, vasti (sum of vastus medialis, lateralis and intermedius); ADD, adductors (sum of adductor brevis, longus and magnus); GMAX, gluteus maximus; GMED, gluteus medius; GAS, gastrocnemius (sum of gastrocnemius medialis and lateralis); HAM, hamstrings (sum of biceps femoris long head, semimembranosus and semitendinosus).

the primary vertical support muscles from ~10% to ~75% of stance, producing up to 1091 N and 704 N of vertical force, respectively. The gastrocnemius produced up to 548 N at ~80% of stance, and declined thereafter along with the contributions from the vasti and soleus.

3.3. Mediolateral redirection

The net GRF was medially-directed throughout stance, indicating acceleration of the centre-of-mass towards the desired cutting direction (Fig. 2C). Medial acceleration of the centre-of-mass was generated primarily by the vasti, gluteus maximus, and gluteus medius. The vasti produced up to 424 N of the medially-directed GRF, peaking at ~75% of stance. Also, the gluteus maximus and medius respectively produced up to 105 N and 96 N of the medially-directed GRF. The gastrocnemius and adductors generated a laterally-directed GRF (thereby opposing acceleration of the centre-of-mass in the direction of travel), with these muscles producing up to 147 N and 102 N, respectively, during late stance. In the last 15% of stance, both the gastrocnemius and soleus accelerated the centre-of-mass medially, although these contributions were no greater than 55 N and 15 N, respectively.

3.4. Muscle contributions to lower-limb joint moments

Overall, the major contributors to the GRFs were also the major contributors to the net joint moments (Fig. 3). Muscles that contributed to propulsion tended to contribute to the hip extension, knee flexion or ankle plantar flexion moments. The exception was soleus, which was the dominant contributor to the ankle plantar flexion moment during the first half of stance (Fig. 3E), but it contributed to braking (posteriorly-directed GRF) during this period (Fig. 2A). The muscles that were responsible for vertical support tended to be the major contributors to the hip extension, knee extension, or ankle plantar flexion moments. The biarticular hamstrings, however, provided an appreciable contribution to the hip extension moment (Fig. 3A), but were responsible more so for generating propulsion rather than vertical support (Fig. 2B). The gluteus medius and maximus were the dominant contributors to the hip abductor moment, whereas the hip adductors contributed to the hip adductor moment at the start and end of stance (Fig. 3B).

4. Discussion

The purpose of this study was to evaluate lower-limb muscle function during an unanticipated sidestep cut. Our main findings

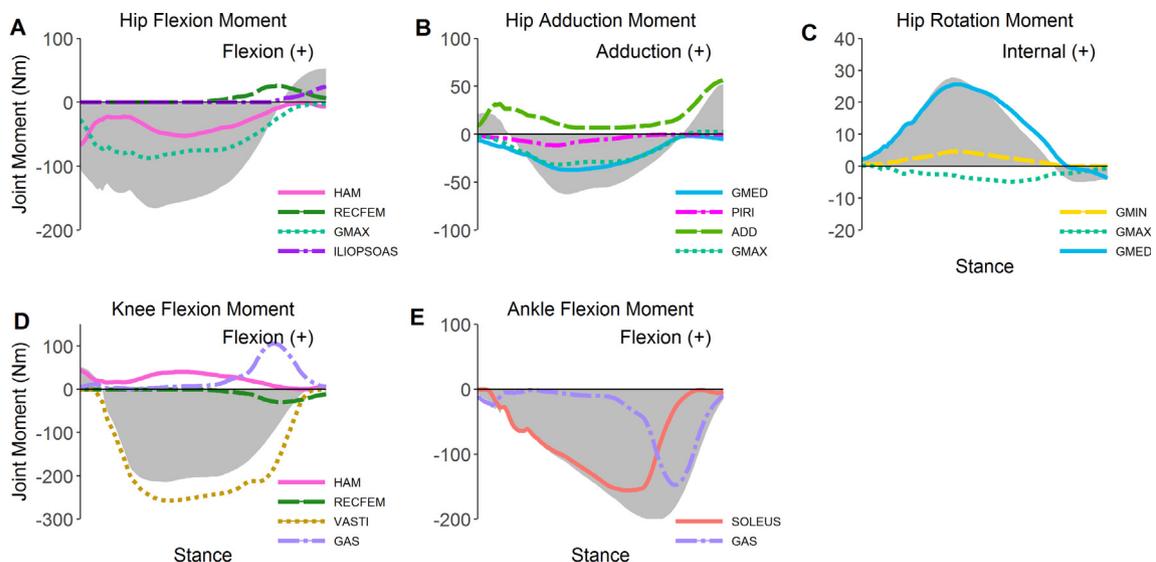


Fig. 3. Muscular contributions to the lower-limb net joint moments for the stance phase of a 45-degree unanticipated sidestep cut. Note that the shaded grey represents the net experimental value determined via inverse dynamics. HAM, hamstrings (sum of biceps femoris long head, semimembranosus and semitendinosus); RECFEM, rectus femoris, GMAX, gluteus maximus; ILIOPSOAS, iliopsoas (sum of iliacus and psoas major); GMED, gluteus medius; PIRI, piriformis; ADD, adductors (sum of adductor brevis, longus and magnus); GMIN, gluteus minimus; VASTI, vasti (sum of vastus medialis, lateralis and intermedius); GAS, gastrocnemius (sum of gastrocnemius medialis and lateralis); SOLEUS, soleus.

were as follows: firstly, braking and propulsion as well as vertical support during a sidestep cut were primarily modulated by the vasti, gluteus maximus, soleus, and gastrocnemius muscles; and secondly, by contributing to the medial GRF, the vasti, gluteus maximus and gluteus medius were the most important muscles for accelerating the centre-of-mass towards the desired cutting direction.

4.1. Vertical support

To our knowledge, no previous studies have used computational musculoskeletal modelling to quantify the contributions of individual lower-limb muscles to the GRFs during sidestep cutting. We have therefore compared our data with prior studies investigating walking (Anderson and Pandy, 2003; John et al., 2012; Lim et al., 2013; Liu et al., 2006; Neptune et al., 2004; Pandy et al., 2010) and running (Dorn et al., 2012b; Hamner and Delp, 2013) in a straight line as well as turning gait (Dixon et al., 2015; Ventura et al., 2015). For example, these studies have consistently found that vertical support is generated by the vasti and gluteus maximus during early stance, after which the gastrocnemius and soleus become the dominant contributors to vertical support. Our observations were similar (Fig. 2B), although we found that the relative contributions of these muscle groups were different during sidestep cutting. Specifically, we found that the vasti made relatively larger contributions to vertical support than the ankle plantar flexors, whereas the opposite has typically been reported for walking (Anderson and Pandy, 2003; Lim et al., 2013; Liu et al., 2006; Neptune et al., 2004) and running (Dorn et al., 2012b; Hamner and Delp, 2013). These observed differences could reflect the specificity of muscle function during sidestep cutting. For example, the sidestep cut requires a larger medially-directed GRF compared to walking (John et al., 2012) and running (Hamner et al., 2013). Since the soleus generated a laterally-directed GRF for the majority of stance (Fig. 2C), there may have been an overall shift in strategy towards greater reliance on the vasti, which produced a medially-directed GRF (Fig. 2C). This explanation is supported by the similar peak soleus muscles forces but substantially higher peak vasti muscle forces in the present study (Supplementary Fig. S4) when compared to running (Dorn et al., 2012b). Nevertheless, we acknowledge that these differences could also be attributable to specific modelling techniques. For example, Dorn and colleagues (Dorn et al., 2012a) found that certain foot-ground contact models predict greater contributions to vertical support from the vasti compared to the soleus during running, whilst other ground-contact models predict the opposite.

4.2. Mediolateral redirection

We found that the vasti, gluteus maximus and gluteus medius were the dominant contributors to redirecting the centre-of-mass acceleration towards the direction of travel in the frontal plane via their contribution to the medially-directed GRF (Fig. 2C). In contrast, the ankle plantar flexors and the adductors were primarily responsible for accelerating the centre-of-mass in the opposite direction (Fig. 2C). Whilst these functional roles are mostly similar to previous investigations of walking (John et al., 2012; Lim et al., 2013; Pandy et al., 2010), our observation that the vasti contributes to a medially-directed GRF is in contrast to prior literature for walking (John et al., 2012; Lim et al., 2013; Pandy et al., 2010) and turning gait (Ventura et al., 2015). The way in which a muscle force results in segment accelerations is dependent on the configuration of the various joints in the system (Zajac and Gordon, 1989). Hence, the differing segmental orientations (i.e. whole body kinematics) between sidestep cutting compared to walking and running (Hamner et al., 2010; Liu et al., 2008) may explain the dif-

fering roles between these locomotor tasks. For example, the stance leg was abducted during sidestep cutting (Supplementary Fig. S1), whereas the stance leg is adducted during walking (Liu et al., 2008). These contrasting limb orientations could redirect the net contribution from the vasti to involve acceleration of the centre-of-mass vertically and medially during sidestep cutting compared to vertically and laterally during walking (John et al., 2012; Lim et al., 2013; Pandy et al., 2010).

4.3. Braking and propulsion

We found that braking (posteriorly-directed GRF) was primarily generated by the vasti throughout stance and by the soleus during the first ~60% of stance, whereas propulsion (anteriorly-directed GRF) was primarily generated by the gluteus maximus and hamstrings during the first 60–70% of stance, and the gastrocnemius and soleus during late stance. Whilst these findings are generally consistent with that for other forms of locomotion (Hamner and Delp, 2013; Lim et al., 2013; Liu et al., 2006, 2008; Neptune et al., 2004), previous studies have typically found that the gluteus maximus primarily accelerates the centre-of-mass posteriorly. As previously discussed, this contrasting result is probably attributable to differing segmental orientations (Zajac and Gordon, 1989) in sidestep cutting compared to walking and running (Hamner and Delp, 2013; Liu et al., 2008).

4.4. Sidestep cutting performance

Determining the specific role of each muscle in the coordination of a rapid sidestep cut may have implications for sidestep cutting performance. For example, prior studies have shown that a greater peak ankle plantar flexor moment (Havens and Sigward, 2015; Marshall et al., 2014) and peak ankle power generation (Marshall et al., 2014) are associated with faster sidestep cutting times. Our data concur with these results, as we found that the main ankle plantar flexors (gastrocnemius and soleus) played a critical role in vertical support, and were the dominant generators of propulsion during late stance (Fig. 2A). Havens and Sigward (Havens and Sigward, 2015) also found that greater sagittal hip power generation was associated with faster sidestep cuts. Our data also support the importance of the hip extensors (gluteus maximus and hamstrings), as these muscles were main generators of propulsion during the first 60% of stance. Additionally, the gluteus maximus was also found to contribute to vertical support as well as acceleration of the centre-of-mass towards the desired cutting direction, demonstrating the importance of this muscle for sidestep cutting performance.

4.5. Implications

Our data may also help inform neuromuscular training interventions, which aim to minimise injury risk and maximise performance. For example, rapid change-in-direction tasks are a common mechanism of ACL injury (Boden et al., 2000; Koga et al., 2010; Olsen et al., 2004) and previous studies have shown that quadriceps and hamstring muscle forces tend to load and unload the ACL, respectively. Our data show that the vasti play a fundamental role in supporting bodyweight and accelerating the centre-of-mass towards the desired cutting direction. This finding suggests that maximising quadriceps force production may be critical for optimal performance, but it could also expose the ACL to greater load. In addition to their role in protecting the ACL from large quadriceps forces (Li et al., 1999), our data suggest that the hamstrings contribute to propulsion (anteriorly-directed GRF) during the first two thirds of stance. It is therefore possible that facilitating hamstrings function during sidestep cutting has the benefit of

minimising injury risk without impairing performance. Further research, however, is needed to verify this assertion.

4.6. Limitations

The present study involved a cohort of eight healthy recreationally-active males. Further research should consider the influence of different populations such as females, specific athletic subgroups, and people with pathology. Additionally, only 45-degree sidestep cutting was investigated. It is possible that greater or smaller cutting angles would alter muscle coordination strategies, which therefore warrants further study.

Muscle forces in the present study were estimated using a static optimisation approach, which does have some limitations. Unfortunately, muscle forces cannot be directly validated because in-vivo muscle forces are not practically feasible to measure (Pandy and Andriacchi, 2010), thus we have no way of directly validating our model predictions. Static optimisation has been shown to be capable of providing accurate predictions of in-vivo joint contact forces at least for walking (Lerner et al., 2015; Wesseling et al., 2015), which serves as an indirect validation due to the high dependency of joint contact forces on muscle forces (Pandy and Andriacchi, 2010). Furthermore, our predicted muscle activations showed reasonable agreement with experimentally recorded EMG data across the stance phase (Fig. 1). It has been suggested that static optimisation may not adequately predict co-contraction of muscles. However, our predicted muscle activations, as well as recently published data (Mokhtarzadeh et al., 2014), do provide evidence of co-contraction. Nevertheless, we recognise that these co-contraction patterns were not necessarily participant-specific, but we do not believe this limitation influenced our conclusions. Further research utilising alternative modelling approaches, such as EMG-driven (Pizzolato et al., 2015) and EMG-hybrid (Sartori et al., 2014) models, may yield further insight.

We also acknowledge that the GRF decomposition technique relies on a ground-contact model that is susceptible to modelling errors (Lin et al., 2011) and can influence the interpretation of muscle function (Dorn et al., 2012a). However, the ground contact model implemented in the present study allowed for a foot-phase-specific constraint set that is more likely to be robust against various foot-strike patterns (Lin et al., 2011). Additionally, verification of our outcomes via the principle of superposition indicated that our model adequately reproduced experimentally measured GRFs, providing further confidence in the suitability of the ground contact model.

Finally, our study has opted to report muscle contributions to GRFs in the global reference frame. This approach facilitates comparison with prior work that has investigated walking (Lim et al., 2013, 2017; Liu et al., 2006; Pandey et al., 2010), running (Dorn et al., 2012b; Hamner and Delp, 2013; Hamner et al., 2010), and stair ambulation (Lin et al., 2015). Unlike these prior studies, the sidestep cutting task reported in the present study has non-planar requirements in order to achieve the 45-degree change in the primary direction of travel, which has the potential to introduce some ambiguity when interpreting our results. The advantage of expressing muscle contributions to GRFs in the global reference frame is that data are presented relative to the initial movement direction, and thus facilitates ease of interpretation and comparison to prior work. A similar approach has been used by another study investigating 90-degree turning gait (Dixon et al., 2015). To evaluate this decision, we plotted the axial rotation angle of the pelvis segment with respect to the global reference frame. We observed that the orientation of the pelvis in the transverse plane did not change substantially throughout stance phase of interest (Supplementary Fig. 1), thus suggesting that the participants' primary direction of travel did not deviate appreciably away from

the anteroposterior axis of the global reference frame. Nonetheless, we have also reported the muscle contributions to the net medio-lateral and anteroposterior GRF (see Supplementary Fig. 6). Inspection of these data may allow readers to visualize how the major muscles contribute to GRFs where the requirements of the task have been taken into account.

5. Conclusion

In summary, this study investigated lower-limb muscle function during a rapid sidestep cut. We found that the vasti, gluteus maximus, soleus, gastrocnemius and hamstrings were important for modulating anteroposterior progression during the stance phase of an unanticipated sidestep cut. These same muscles (except the hamstrings) were also important for supporting body-weight, while the vasti, gluteus maximus and the gluteus medius played a critical role in accelerating the centre-of-mass towards the desired cutting direction. These functional roles should be considered in neuromuscular retraining programs which aim to reduce injury risk and/or maximise performance.

Author contributions

Conception of experimental procedures – NM, AGS & DAO. Conception of data analysis – NM & AGS. Data collection and analysis – NM. Preparation of Figures – NM. Interpretation of data – NM, AGS, MHC & DAO. Writing of manuscript – NM, AGS, MHC & DAO.

Conflicts

Authors have no conflicts of interest to declare.

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Appendix A. Supplementary material

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.jbiomech.2018.10.021>.

References

- Anderson, F.C., Pandey, M.G., 2003. Individual muscle contributions to support in normal walking. *Gait Posture* 17, 159–169.
- Beaulieu, M.L., Lamontagne, M., Xu, L., 2009. Lower limb muscle activity and kinematics of an unanticipated cutting manoeuvre: a gender comparison. *Knee Surg. Sport. Tr. A* 17, 968–976.
- Benoit, D.L., Ramsey, D.K., Lamontagne, M., Xu, L., Wretenberg, P., Renström, P., 2006. Effect of skin movement artifact on knee kinematics during gait and cutting motions measured in vivo. *Gait Posture* 24, 152–164.
- Bloomfield, J., Polman, R., O'Donoghue, P., 2007. Physical demands of different positions in FA premier league soccer. *J. Sports Sci. Med.* 6, 63–70.
- Boden, B.P., Dean, G.S., Feagin Jr, J.A., Garrett Jr, W.E., 2000. Mechanisms of anterior cruciate ligament injury. *Orthopedics* 23, 573–578.
- Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans. Bio-med. Eng.* 54, 1940–1950.
- Dixon, P.C., Jansen, K., Jonkers, I., Stebbins, J., Theologis, T., Zavatsky, A.B., 2015. Muscle contributions to centre of mass acceleration during turning gait in typically developing children: a simulation study. *J. Biomech.* 48, 4238–4245.
- Dorn, T.W., Lin, Y.-C., Pandey, M.G., 2012a. Estimates of muscle function in human gait depend on how foot-ground contact is modelled. *Comput. Meth. Biomech. Biomed. Eng.* 15, 657–668.

- Dorn, T.W., Schache, A.G., Pandy, M.G., 2012b. Muscular strategy shift in human running: dependence of running speed on hip and ankle muscle performance. *J. Exp. Biol.* 215, 1944–1956.
- Green, B.S., Blake, C., Caulfield, B.M., 2011. A comparison of cutting technique performance in rugby union players. *J. Strength Cond. Res.* 25, 2668–2680.
- Hamner, S.R., Delp, S.L., 2013. Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds. *J. Biomech.* 46, 780–787.
- Hamner, S.R., Seth, A., Delp, S.L., 2010. Muscle contributions to propulsion and support during running. *J. Biomech.* 43, 2709–2716.
- Hamner, S.R., Seth, A., Steele, K.M., Delp, S.L., 2013. A rolling constraint reproduces ground reaction forces and moments in dynamic simulations of walking, running, and crouch gait. *J. Biomech.* 46, 1772–1776.
- Havens, K.L., Sigward, S.M., 2015. Cutting mechanics: relation to performance and anterior cruciate ligament injury risk. *Med. Sci. Sports Exerc.* 47, 818–824.
- Hicks, J.L., Uchida, T.K., Seth, A., Rajagopal, A., Delp, S.L., 2015. Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of movement. *J. Biomech. Eng.* 137, 020905.
- John, C.T., Seth, A., Schwartz, M.H., Delp, S.L., 2012. Contributions of muscles to mediolateral ground reaction force over a range of walking speeds. *J. Biomech.* 45, 2438–2443.
- Karcher, C., Buchheit, M., 2014. On-court demands of elite handball, with special reference to playing positions. *Sports Med. (Auckland, N.Z.)* 44, 797–814.
- Koga, H., Nakamae, A., Shima, Y., Iwasa, J., Myklebust, G., Engebretsen, L., Bahr, R., Krosshaug, T., 2010. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am. J. Sports Med.* 38, 2218–2225.
- Kristianslund, E., Krosshaug, T., Van den Bogert, A.J., 2012. Effect of low pass filtering on joint moments from inverse dynamics: implications for injury prevention. *J. Biomech.* 45, 666–671.
- Lerner, Z.F., DeMers, M.S., Delp, S.L., Browning, R.C., 2015. How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces. *J. Biomech.* 48, 644–650.
- Li, G., Rudy, T.W., Sakane, M., Kanamori, A., Ma, C.B., Woo, S.L., 1999. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J. Biomech.* 32, 395–400.
- Lim, Y.P., Lin, Y.-C., Pandy, M.G., 2013. Muscle function during gait is invariant to age when walking speed is controlled. *Gait Posture* 38, 253–259.
- Lim, Y.P., Lin, Y.-C., Pandy, M.G., 2017. Effects of step length and step frequency on lower-limb muscle function in human gait. *J. Biomech.* 57, 1–7.
- Lin, Y.-C., Fok, L.A., Schache, A.G., Pandy, M.G., 2015. Muscle coordination of support, progression and balance during stair ambulation. *J. Biomech.* 48, 340–347.
- Lin, Y.C., Kim, H.J., Pandy, M.G., 2011. A computationally efficient method for assessing muscle function during human locomotion. *Int. J. Numer. Meth. Biomed. Eng.* 27, 436–449.
- Liu, M.Q., Anderson, F.C., Pandy, M.G., Delp, S.L., 2006. Muscles that support the body also modulate forward progression during walking. *J. Biomech.* 39, 2623–2630.
- Liu, M.Q., Anderson, F.C., Schwartz, M.H., Delp, S.L., 2008. Muscle contributions to support and progression over a range of walking speeds. *J. Biomech.* 41, 3243–3252.
- Lu, T.-W., O'connor, J., 1999. Bone position estimation from skin marker coordinates using global optimisation with joint constraints. *J. Biomech.* 32, 129–134.
- Maniar, N., Schache, A.G., Sritharan, P., Opar, D.A., 2018. Non-knee-spanning muscles contribute to tibiofemoral shear as well as valgus and rotational joint reaction moments during unanticipated sidestep cutting. *Sci. Rep.* 8, 2501.
- Marshall, B.M., Franklyn-Miller, A.D., King, E.A., Moran, K.A., Strike, S.C., Falvey, E.C., 2014. Biomechanical factors associated with time to complete a change of direction cutting maneuver. *J. Strength Cond. Res.* 28, 2845–2851.
- Millard, M., Uchida, T., Seth, A., Delp, S.L., 2013. Flexing computational muscle: modeling and simulation of musculotendon dynamics. *J. Biomech. Eng.* 135, 021005.
- Mokhtarzadeh, H., Perraton, L., Fok, L., Munoz, M.A., Clark, R., Pivonka, P., Bryant, A.L., 2014. A comparison of optimisation methods and knee joint degrees of freedom on muscle force predictions during single-leg hop landings. *J. Biomech.* 47, 2863–2868.
- Mokhtarzadeh, H., Yeow, C.H., Hong Goh, J.C., Oetomo, D., Malekipoor, F., Lee, P.V.-S., 2013. Contributions of the Soleus and Gastrocnemius muscles to the anterior cruciate ligament loading during single-leg landing. *J. Biomech.* 46, 1913–1920.
- Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. *J. Biomech.* 34, 1387–1398.
- Neptune, R.R., Wright, I.C., van den Bogert, A.J., 1999. Muscle coordination and function during cutting movements. *Med. Sci. Sports Exerc.* 31, 294–302.
- Neptune, R.R., Zajac, F.E., Kautz, S.A., 2004. Muscle force redistributes segmental power for body progression during walking. *Gait Posture* 19, 194–205.
- Oliveira, A., Silva, P., Lund, M.E., Kersting, U.G., Farina, D., 2013. Fast changes in direction during human locomotion are executed by impulsive activation of motor modules. *Neuroscience* 228, 283–293.
- Olsen, O.-E., Myklebust, G., Engebretsen, L., Bahr, R., 2004. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am. J. Sports Med.* 32, 1002–1012.
- Pandy, M.G., Andriacchi, T.P., 2010. Muscle and joint function in human locomotion. *Annu. Rev. Biomed. Eng.* 12, 401–433.
- Pandy, M.G., Lin, Y.-C., Kim, H.J., 2010. Muscle coordination of mediolateral balance in normal walking. *J. Biomech.* 43, 2055–2064.
- Pizzolato, C., Lloyd, D.G., Sartori, M., Ceseracciu, E., Besier, T.F., Fregly, B.J., Reggiani, M., 2015. CEINMS: a toolbox to investigate the influence of different neural control solutions on the prediction of muscle excitation and joint moments during dynamic motor tasks. *J. Biomech.* 48, 3929–3936.
- Rajagopal, A., Dembia, C.L., DeMers, M.S., Delp, D.D., Hicks, J.L., Delp, S.L., 2016. Full-body musculoskeletal model for muscle-driven simulation of human gait. *IEEE Trans. Bio-med. Eng.* 63, 2068–2079.
- Samaan, M.A., Weinhandl, J.T., Bawab, S.Y., Ringleb, S.I., 2016. Determining residual reduction algorithm kinematic tracking weights for a sidestep cut via numerical optimization. *Comput. Meth. Biomech. Biomed. Eng.* 19, 1721–1729.
- Sartori, M., Farina, D., Lloyd, D.G., 2014. Hybrid neuromusculoskeletal modeling to best track joint moments using a balance between muscle excitations derived from electromyograms and optimization. *J. Biomech.* 47, 3613–3621.
- Sigward, S.M., Powers, C.M., 2006. The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. *Clin. Biomech. (Bristol, Avon)* 21, 41–48.
- Sritharan, P., Lin, Y.C., Pandy, M.G., 2012. Muscles that do not cross the knee contribute to the knee adduction moment and tibiofemoral compartment loading during gait. *J. Orthopaed. Res.: Off. Publ. Orthopaed. Res. Soc.* 30, 1586–1595.
- Ventura, J.D., Klute, G.K., Neptune, R.R., 2015. Individual muscle contributions to circular turning mechanics. *J. Biomech.* 48, 1067–1074.
- Walker, P.S., Rovick, J.S., Robertson, D.D., 1988. The effects of knee brace hinge design and placement on joint mechanics. *J. Biomech.* 21, 965–974.
- Weinhandl, J.T., Earl-Boehm, J.E., Ebersole, K.T., Huddleston, W.E., Armstrong, B.S., O'Connor, K.M., 2013. Anticipatory effects on anterior cruciate ligament loading during sidestep cutting. *Clin. Biomech. (Bristol, Avon)* 28, 655–663.
- Wesseling, M., Derikx, L.C., De Groot, F., Bartels, W., Meyer, C., Verdonschot, N., Jonkers, I., 2015. Muscle optimization techniques impact the magnitude of calculated hip joint contact forces. *J. Orthopaed. Res.: Off. Publ. Orthopaed. Res. Soc.* 33, 430–438.
- Zajac, F.E., Gordon, M.E., 1989. Determining muscle's force and action in multi-articular movement. *Exerc. Sport Sci. Rev.* 17, 187–230.
- Zajac, F.E., Neptune, R.R., Kautz, S.A., 2002. Biomechanics and muscle coordination of human walking: part I: introduction to concepts, power transfer, dynamics and simulations. *Gait Posture* 16, 215–232.