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Optimizing whole-body kinematics to minimize valgus knee loading during sidestepping: Implications for ACL injury risk

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ABSTRACT

The kinematic mechanisms associated with elevated externally applied valgus knee moments during non-contact sidestepping and subsequent anterior cruciate ligament (ACL) injury risk are not well understood. To address this issue, the residual reduction algorithm (RRA) in OpenSim was used to create nine subject-specific, full-body (37 degrees of freedom) torque-driven simulations of athletic males performing unplanned sidestep (UnSS) sport tasks. The RRA was used again to produce an optimized kinematic solution with reduced peak valgus knee torques during the weight acceptance phase of stance. Pre-to-post kinematic optimization, mean peak valgus knee moments were significantly reduced by 44.2 Nm ($p=0.045$). Nine of a possible 37 upper and lower body kinematic changes in all three planes of motion were consistently used during the RRA to decrease peak valgus knee moments. The generalized kinematic strategy used by all nine simulations to reduce peak valgus knee moments and subsequent ACL injury risk during UnSS was to redirect the whole-body center of mass medially, towards the desired direction of travel.

1. Introduction

Anterior cruciate ligament (ACL) injuries in sport are common (Gianotti et al., 2009; Janssen et al., 2011). New Zealand and Australia spend approximately 17.4 million NZD (Gianotti et al., 2009) and 75 million AUD (Janssen et al., 2011) on ACL injuries each year. Extrapolating from figures reported by Gianotti et al. (2009) and current world population estimates (World Bank, 2010); the United States annually spend approximately 1 billion USD on ACL injury management. Approximately 55% of ACL injured athletes are not capable of returning to the same level of competition two years post-reconstruction (Dunn and Spindler, 2010), a percent that increases to 70% after three years (Roos et al., 1995), which were over double that of a comparable group of non-ACL injured athletes (Ekstand et al., 1990; Roos et al., 1995). A rupture to the ACL can be considered one of the most severe knee injuries an athlete can sustain in sport.

More than one half of non-contact ACL injuries occur during sidestepping sport manoeuvres (Cochrane et al., 2007; Koga et al., 2010; Krosshaug et al., 2007). Biomechanical studies have shown that during the weight acceptance (WA) phase of sidestepping, which is from initial heel contact to the first trough in the vertical ground reaction force vector (Dempsey et al., 2007), peak valgus knee moments are up to 2-times larger than those observed during straight-line running (Besier et al., 2001). During weight-bearing (i.e. stance) (Fleming et al., 2001) and when valgus knee moments are combined with anterior tibial translations, ACL strain is significantly elevated (Markolf et al., 1995; Withrow et al., 2006). These are similar to the loading patterns needed to increase ACL strain and/or reach injurious loading thresholds *in-silico* (McLean et al., 2004; McLean et al., 2008; Quatman et al., 2011; Shin et al., 2011). Reducing valgus knee loading during sport tasks like sidestepping is therefore considered an appropriate countermeasure to reduce ACL injury risk.

Hewett et al. (2005) has shown peak valgus knee moments during landing are good predictors of ACL injury. Peak valgus knee moments (Besier et al., 2001; Chaudhari et al., 2005; Dempsey et al., 2007; McLean et al., 2005) and peak in-vivo ACL strain (Cerulli et al., 2003) are generally observed during WA. Consequently, one focus of ACL injury prevention training intervention is to reduce valgus knee moments during the WA phase of sidestepping (Cochrane et al., 2010; Dempsey et al., 2009), when ACL injury risk is thought to be the greatest.

Both neuromuscular (Myer et al., 2005) and balance (Cochrane et al., 2010) training have been shown to reduce valgus knee moments during landing and sidestepping. However, these studies have not measured and/or identified the kinematic mechanisms

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contributing to these observed reductions in knee loading. Hip (McLean et al., 2005), trunk (Dempsey et al., 2007) and arm kinematics (Chaudhari et al., 2005) have been shown to be associated with peak valgus knee moments during sidestepping, while lateral trunk stability has been shown to be associated with rate of ACL injury (Zazulak et al., 2007). Although associations between upper body biomechanics and knee loading have been identified, they are heuristic in nature, providing limited causal information when applied to complex, multi-body, dynamic movements like sidestepping.

Full-body *in-silico* simulations, with optimization computational methods have been used previously to identify causal relationships between whole-body (WB) kinematics and peak varus knee moments during walking (Fregly et al., 2007). The open-source musculoskeletal modeling software OpenSim (simtk.org, Stanford, CA) allows for *in-silico* simulations of human movement to be created from three-dimensional (3D) motion data. The residual reduction algorithm (RRA) within OpenSim is an optimization tool capable of altering a simulation's kinematics to reduce peak knee joint loading during sidestepping. Using this modeling framework and these computational tools, our aim was to identify causal relationships between WB kinematics and peak valgus knee moments during the WA phase of sidestepping.

2. Methods

The experimental procedure consisted of three phases: (1) experimental motion data collection; (2) skeletal modeling and residual force/moment reduction; and (3) minimizing peak valgus knee torques by optimizing WB kinematics (Fig. 1).

Thirty-four male Western Australian Amateur Football players completed the UWA sidestepping protocol at 5 ms^{-1} (Besier et al., 2001; Dempsey et al., 2007). All experimental procedures were approved by the University of Western Australia Human Research Ethics Committee and all participants provided their informed written consent prior to data collection. WB kinematics and ground reaction forces (GRF) were recorded from a series of straight-line runs, together with pre-planned and unplanned (UnSS) sidestep trials as described in Dempsey et al. (2007). Inverse dynamics (ID) was used to calculate peak valgus knee moments during the WA phase of sidestepping. From this cohort, nine participants with the largest mean peak valgus knee moments, which always occurred during UnSS, were chosen for further analysis. The nine participants were 22.0 ± 4.3 years of age, with a mean height and body mass of 1.83 ± 0.04 m and 80.8 ± 6.66 kg, respectively.

A 12-camera 250 Hz VICON MX motion capture system (VICON Peak, Oxford Metrics Ltd., UK) recorded 3D full-body kinematics (Dempsey et al., 2007). GRF were synchronously recorded at 2,000 Hz from a single 1.2×1.2-m force platform (Advanced Mechanical Technology Inc., Watertown, MA.). Kinematic and GRF data were both low pass filtered at 15 Hz using a zero-lag 4th order Butterworth digital filter in Workstation (Vicon Peak, Oxford Metrics Ltd., UK). The cut-off frequency was selected based on a residual analysis (Winter, 2005) and visual inspection. Applying the same filter and cut-off frequency to the motion and GRF data has been shown to reduce knee joint kinetic artifacts (Bisseling and Hof, 2006).

Custom biomechanical models in Matlab (Matlab 7.8, The Math Works, Inc., Natick, Massachusetts, USA), Vicon Bodybuilder (Dempsey et al., 2007) and functional knee and hip joint methods (Besier et al., 2003) were used to calculate subject-specific joint centers. Joint centers, marker trajectories and GRF data were then exported into OpenSim 1.9.1.

A 14 segment, 37 degree-of-freedom (DoF) rigid-linked skeletal models driven by 37 ideal torque actuators formed the foundation of each simulation (see supplementary material). For clarity, joint torques, not muscles were used to drive each simulation. Twenty-nine of the model's DoF have been described previously (Hamner et al., 2010), to which we added 2 DoF wrist joints (flexion/extension and ulnar/radial deviation) and 3 DoF knee joints (flexion/extension, internal/external rotation, and varus/valgus). Internal/external rotation and the varus/valgus DoF of the knee were modeled as universal joints, with the same center of rotation, and moved with the flexion/extension DoF, which was modeled as a planar joint, allowing the tibia to translate relative to the femur as a function of knee flexion angle (Delp et al., 1990) (Fig. 2). Segment lengths were scaled to each participant's subject-specific joint center positions, where segment masses and inertial properties were scaled to each participant's total body mass in OpenSim.

Inverse kinematics (IK) (Delp et al., 2007) is a global optimization method (weighted least-squares) used in OpenSim to calculate a model's generalized coordinates (i.e. joint angles) during the WA phase of UnSS. This is done by minimizing the squared distances between the rigid segment markers of the 37 DoF rigid-linked skeletal model and the experimentally recorded kinematics by adjusting the model's generalized coordinates. Following IK, the generalized coordinates and experimental GRF measures were used in a two-step RRA process within OpenSim.

Inconsistencies between a model's dynamics and experimental GRF measures ($\sum F_{model} \neq GRF$) called residual forces and moments are often overlooked when using ID (Delp et al., 2007). Residual forces and moments, which are present in all biomechanical models, represent errors and assumptions in the modeling process (i.e. joint center and inertial estimates). OpenSim addresses this issue by creating a 6 DoF joint between the pelvis and ground, holding residual forces and moments not solved for during ID, satisfying Newton's second law ($\sum F_{model} + \sum F_{residuals} = GRF$). The goal of the RRA is to produce a set of actuator forces (i.e. joint torques) to generate joint motions that track a desired set of generalized coordinates, while minimizing the model's residual forces and moments (Delp et al., 2007; Thelen and Anderson, 2006). The result is simulation that tracks the experimentally recorded GRF with dynamic consistency.

The first step of RRA optimizes trunk center of mass (CoM) position to reduce mean residual force and moment offsets. The second step reduces these residuals further by slightly adjusting the model's generalized coordinates by minimizing the sum of three components ($J(x)$): (1) the weighted ($w_{\ddot{q}_i}$) squared errors between the experimental ($\ddot{q}_i^{exp.}$) and simulated ($\ddot{q}_i^{sim.}$) accelerations for each DoF (n_q) of the skeletal model; (2) the squared residual forces/torques (R_j) ($n=6$) proportional to their driving excitation (x_j^R) normalized by their maximum residual forces/torques (R_j^{max}); and (3) the squared joint torques (T_k) for each DoF (n_T) in the model, proportional to their driving excitation (x_k^T) normalized by their maximum torques (T_k^{max})

$$J(x) = \min_{\boldsymbol{x}^R,\boldsymbol{x}^T} \left[\sum_{i=1}^{n_q} w_{\ddot{q}_i} (\ddot{q}_i^{exp.} - \ddot{q}_i^{sim.})^2 + \sum_{j=1}^{6} \left(\frac{R_j(x_j^R)}{R_j^{max}} \right)^2 + \sum_{k=1}^{n_T} \left(\frac{T_k(x_k^T)}{T_k^{max}} \right)^2 \right] \quad (1)$$

The RRA establishes a set of excitations ($\boldsymbol{x}^R$,$\boldsymbol{x}^T$), driving the six residual forces/torques (R_j) and joint torques (T_k) in a simulation. Using numerical integration (i.e. $q^{sim.} = \int\int \ddot{q}_i^{sim.}$), the model's dynamics are generated at each time step. A time varying set of generalized joint coordinates ($q_i^{sim.}$) residual forces/torques (R_j), and joint torques (T_k) are output, producing a dynamically consistent torque-driven simulation of UnSS. During RRA, the orientation of the GRF vector is held relative to the stance foot's CoM. Thus the location of the center of pressure is always in the same relative orientation to the stance foot's CoM during the RRA process.

The results of RRA depended on what values are chosen for the 74 (i.e. $74 = n_T + 6 + n_q$) input parameters, which include maximal joint torques, maximal pelvic residual forces/torques and kinematic weightings. To produce the best possible dynamically consistent simulation, input parameters were solved using an outer optimization method, which minimized joint torques (i.e. $\min(T^{sim.})^2$), residual error (i.e. $\min(R^{sim.})^2$) and total kinematic error (i.e. $\min[(q^{exp.} - q^{sim.})^2]$) (Reinbolt et al., 2011). Additional weightings were placed on the residuals and kinematic errors, meaning the primary goal of the external

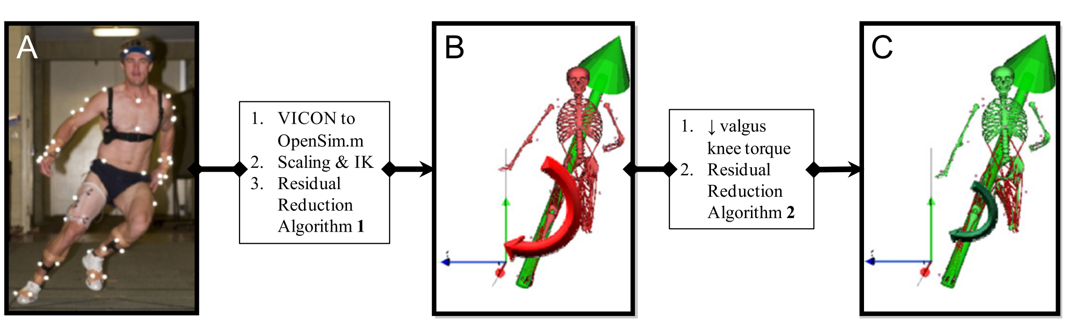

Fig. 1. Overview of the experimental procedure: motion data collection (A), skeletal modeling and residual reduction (B) and optimization WB kinematics to minimized peak valgus knee moments (C).

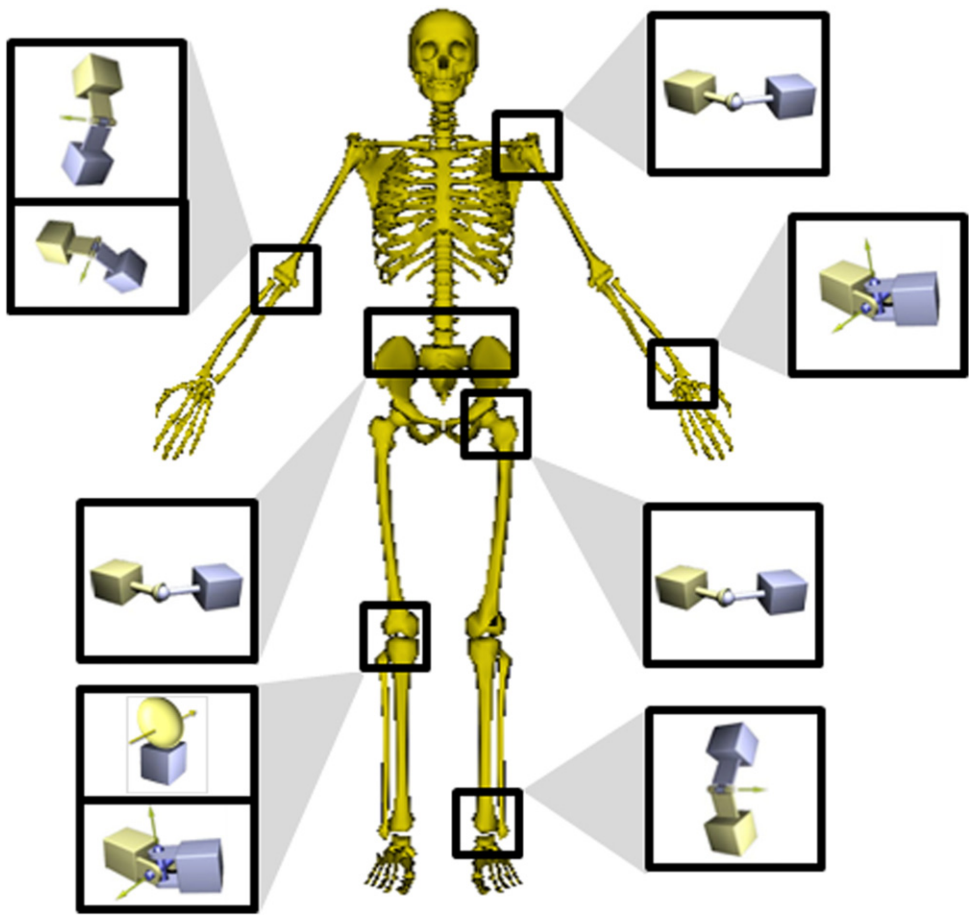

Fig. 2. Depiction of 37 DoF, 14 segment full-body rigid-linked skeletal model. The pelvis segment with respect to ground was defined using 3 translations and 3 rotations (6 DoF). A ball-and-socket was used to represent the hip, shoulder and pelvis to trunk/head joints (3 DoF). The wrists were modeled as universal joints (2 DoF). The radial-ulnar, elbow and ankle joints were modeled as revolutes (1 DoF). The knee joint (3 DoF) was modeled as a planar joint in the flexion/extension axis which allowed the tibia to translate as a function of knee flexion angle (Delp et al., 1990); internal/external rotation and abd/adduction were modeled as universal joints.

optimization method was to minimize residual and kinematic error during RRA. Using these methods, peak residual forces and moments were less than 2.5 N and 0.5 Nm, respectively. Maximum root mean squared joint coordinate errors were between 1.0–8.4°, with a mean of $3.5 \pm 2.8°$ for all nine simulations. Furthermore, in a randomly selected subset of subjects, we compared external knee moment traces produced from RRA with those calculated using ID. Peak knee moments were within $\pm 5\%$, and occurred within ± 0.016 s of each other. Given these results we were confident the simulated UnSS knee moments were consistent with those reported previously in the literature (Besier et al., 2001; Dempsey et al., 2007; 2009).

The final stage of this procedure was to minimize peak valgus knee moments during the WA phase of UnSS. This was accomplished by reducing the maximum joint torque ($T_{V/V}^{max}$) value associated with the knee's V/V DoF and RRA re-run using the same maximal torques, maximal residual forces/torques and kinematic weightings solved for using the external optimization method along with the experimental GRF measures. For a kinematic optimization solution to be deemed acceptable, stance foot translations were limited to 30 mm (Fregly et al., 2007) in all three directions i.e. medial/lateral (M/L), anterior/posterior (A/P) and inferior/superior (I/S). It should be noted additional kinematic constraints were placed on the stance foot to restrict foot translations during the RRA process.

$$K(J(x)) = \min_{x^R,x^T}\left[\sum_{i=1}^{n_q} w_{\ddot{q}}(\ddot{q}_i^{exp.} - \ddot{q}_i^{sim.})^2 + \sum_{j=1}^{6}\left(\frac{R_j(x_j^R)}{R_j^{max}}\right)^2 + \left(\frac{T_{V/V}(x_{V/V}^T)}{T_{V/V}^{max}}\right)^2 + \sum_{k=1}^{n_T-1}\left(\frac{T_k(x_K^T)}{T_k^{max}}\right)^2\right] \tag{2}$$

Pre-to-post kinematic optimization, selected kinematic and kinetic variables were analyzed during WA. Peak valgus, flexion and internal rotation knee torques calculated during RRA were expressed as externally applied knee moments (Lloyd, 2001). The mean difference in WB CoM relative to stance foot CoM as well as relative stance foot CoM orientation pre-to-post kinematic optimization were calculated in the M/L, A/P and I/S directions.

Independent one-way ANOVAs were used to compare peak mean valgus, flexion and internal rotation knee moments pre-to-post kinematic optimization ($\alpha = 0.05$). Independent one-way ANOVAs with a Bonferroni *post hoc* test were used to determine if significant differences in WB CoM relative to stance foot CoM were observed pre-to-post kinematic optimization between the M/L, A/P and I/S directions ($\alpha = 0.05$). The mean M/L, A/P and I/S relative error (%) of the stance foot's CoM trajectory pre-to-post kinematic optimization were also calculated.

The mean angular differences for all 37 DoF pre-to-post kinematic optimization were calculated in 20% intervals over WA, and kinematic maps created for all nine simulations (Fig. 3). Each map represented the absolute change in joint angles pre-to-post kinematic optimization, for each DoF within the skeletal model ($n = 37$), in each of the five time intervals within WA. The mean difference of all joints across all time points during WA was then calculated. Any joint with an angular kinematic change greater than 2σ above the mean was defined as a critical joint coordinate and identified as a kinematic change that most influenced the observed changes in peak valgus knee moments pre-to-post kinematic optimization.

3. Results

Pre-to-post kinematic optimization, peak mean valgus knee moments during UnSS were significantly reduced by 44.2 Nm (106.1 ± 48.6 to 61.9 ± 36.4 Nm) ($p = 0.045$). Peak mean flexion and internal rotation knee moments increased by 24.1 Nm (252.2 ± 80.2 to 276.3 ± 69.4 Nm) and 1.1 Nm (7.6 ± 6.9 to 8.7 ± 7.7 Nm), respectively (Fig. 4).

Pre-to-post kinematic optimization, unique 3D kinematic changes were used by each simulation to reduce peak valgus knee moments. However, only nine of a possible 37 critical joint

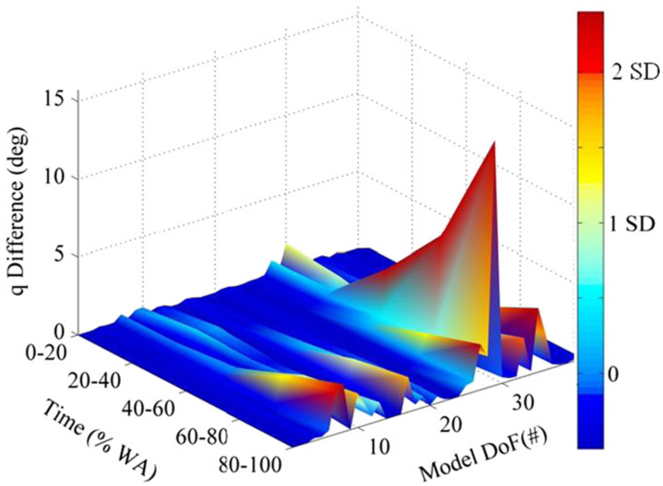

Fig. 3. Kinematic mapping of a typical simulation representing the absolute kinematic changes (q) from pre-to-post kinematic optimization for all DoF within the skeletal model ($n=37$) at 20% intervals during WA of UnSS.

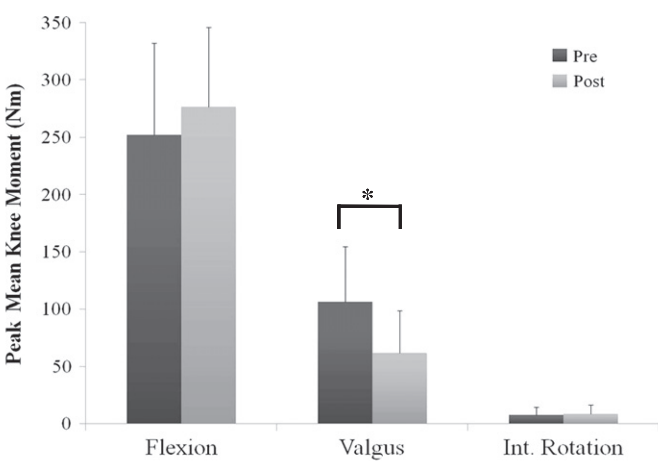

Fig. 4. Peak mean knee flexion, valgus and internal rotation moments pre-to-post kinematic optimization calculated during the WA phase of an UnSS. Symbol * indicates a significant change over time ($\alpha=0.05$).

coordinates were used by all nine simulations to reduce peak valgus knee moments during UnSS (Table 1) (See supplementary material for video data of a typical simulation pre-to-post kinematic optimization). Two primary kinematic strategies were used by the simulations to reduce peak valgus knee moments: The first, used by six of the nine simulations elevated mean ankle plantar flexion by $7.9\pm5.2^{\circ}$, while the second, used by all nine simulations, was to reposition WB CoM medially and anteriorly relative to the stance foot CoM, which was towards desired direction of travel during the UnSS.

Supplementary material related to this article can be found online at doi:10.1016/j.jbiomech.2012.02.010.

The mean change in WB CoM relative to stance foot CoM was 3.1 ± 0.8 cm medially, 1.4 ± 1.4 cm anteriorly and 0.2 ± 0.3 cm superiorly. The mean change in WB CoM was significantly different between the M/L, A/P and I/S directions ($p<0.001$). *Post hoc* analysis showed that mean changes in WB CoM were significantly greater in the medial direction relative to anterior ($p=0.003$) and superior directions ($p<0.001$), while anterior changes were significantly greater than changes in the superior direction ($p=0.045$) (Fig. 5).

Mean changes in stance foot CoM were limited to -3.7 ± 2.0, 10.5 ± 2.97 and -3.1 ± 2.1 mm in the M/L, A/P and I/S directions, respectively. The mean relative error of the stance foot's CoM

Table 1
Individual simulation (Sim), mean (μ) differences of critical joint coordinates (deg) and mean WB CoM relative to stance foot CoM (m) pre-to-post kinematic optimization. Anterior, medial and superior changes in degrees are positive. Anterior and medial are both towards the desired change of direction pathway. The symbol "–" means the variable was not identified as a critical joint coordinate.

	Sagittal Plane (deg)				Frontal Plane (deg)			Transverse Plane (deg)		WB CoM v Foot CoM (cm)		
	R_Plant_Flex	L_Knee_Ext	L_Hip_Flex	L_Shoulder_Ext	R_Shoulder_Ad	L_Hip_Ab	R_Hip_Ab	Trunk_Rot_Med	R_Shoulder_Int_Rot	Ant/Post	Med/Lat	Sup/Inf
Sim 1	-	-	-	4.1	15.7	-	3.1	2.9	-	1.2	4.8	−0.1
Sim 2	9.4	7.1	6.8	-	16.1	-	-	-	-	4.6	3.0	0.1
Sim 3	3.8	-	-	-	9.4	-	-	-	-	0.9	2.3	0.3
Sim 4	5.4	-	-	-	24.9	-	-	-	-	0.8	2.5	0.1
Sim 5	-	-	1.7	-	-	2.0	-	3.2	1.0	2.4	3.6	0.2
Sim 6	16.5	-	-	-	16.2	-	-	-	-	2.0	2.5	1.0
Sim 7	-	6.1	-	8.3	6.2	-	-	-	-	-0.5	2.9	0.0
Sim 8	10.0	2.6	-	-	-	6.6	-	-	-	1.0	3.7	0.3
Sim 9	2.2	-	-	-	5.1	-	-	-	-	0.4	2.9	0.0
μ	7.9	5.3	4.3	6.2	13.4	4.3	3.1	3.1	1.0	1.4	3.1	0.2
σ	5.2	2.4	3.6	3.0	6.9	3.3	-	0.2	-	1.4	0.8	0.3
n	6	3	2	2	7	2	1	2	1	9	9	9

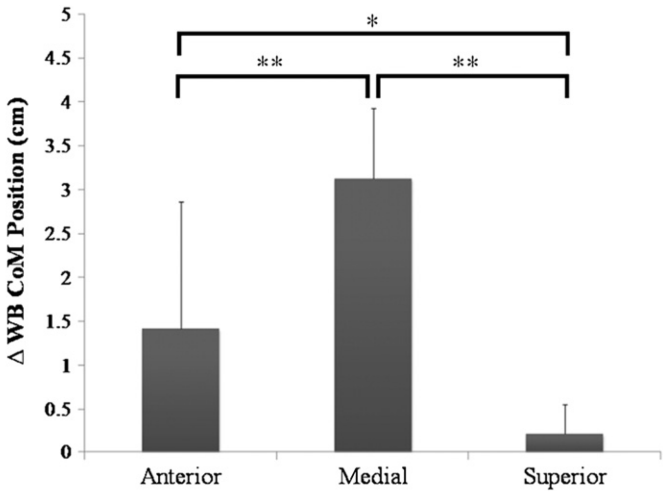

Fig. 5. Mean peak changes in WB CoM relative to stance foot CoM pre-to-post kinematic optimization. Anterior and medial changes are towards the desired change of direction pathway. Symbols * and ** indicated a significant change of $p < 0.05$ and $p < 0.01$, respectively.

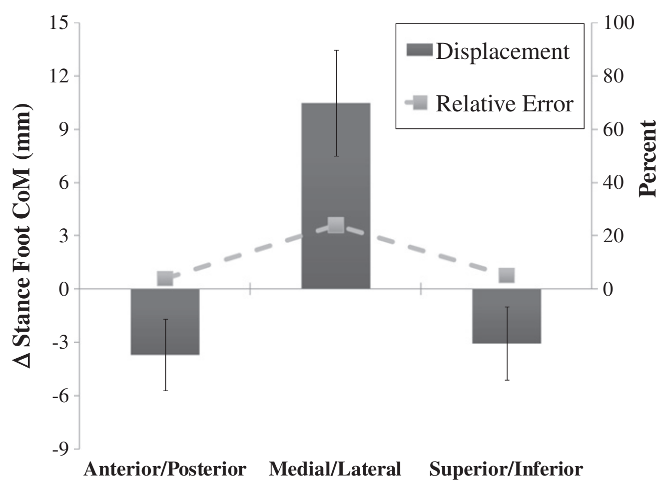

Fig. 6. Mean change in stance foot CoM (mm) and relative error (%) with respect to the original foot trajectory pre-to-post kinematic optimization. Anterior, medial and superior changes are positive.

trajectory was 4.0, 23.9 and 5.1% in the M/L, A/P and I/S directions, respectively (Fig. 6).

4. Discussion

Associations between upper body posture and peak valgus knee moments during sidestepping have been reported previously in the literature (Chaudhari et al., 2005; Dempsey et al., 2007; McLean et al., 2005). For example, lateral trunk flexion (Dempsey et al., 2007) and constraining an athlete's arms to their mid-line (Chaudhari et al., 2005) likely restricted their upper body CoM from moving medially during sidestepping, resulting in the observed increases in peak valgus knee moments. Results from this study confirm that upper body kinematics indeed influence valgus knee loading during sidestepping. However, unlike previous findings, results showed that one kinematic change was always coupled with kinematic changes from at least one other joint along the kinematic chain. Additionally, results showed that both upper and lower body kinematic changes in all three planes of motion can be utilized to decrease peak valgus knee loading during UnSS. The generalized kinematic strategy used by all nine simulations to reduce peak valgus knee moments during UnSS was to reposition WB CoM medially, towards the desired direction of travel.

Statistically significant reductions in peak valgus knee moments were accompanied by increases in both peak flexion and internal rotation knee moments. Increases in flexion knee moments combined with decreases in peak varus knee moments have been observed following gait re-training in clinical settings (Fregly et al., 2007; Walter et al., 2010). In addition, elevated applied flexion moments in isolation are unlikely to reach an injurious loading threshold *in-silico* (McLean et al., 2004), while the observed increases in internal rotation knee moments are considered negligible. Results from this and previous literature suggest that the *in-silico* changes in knee moments are consistent with clinical findings and were effective in reducing surrogate measures of ACL injury risk.

As with all optimization based research, an enormous kinematic solution space exists. As such, unique kinematic strategies were used by each simulation to reduce peak valgus knee moments during UnSS. Though the results showed each simulation consistently used the same nine of a possible 37 joint DoF to reduce peak valgus knee moments during UnSS, 511 (2^9-1) kinematic combinations remain. The experimental and computational time required to process a single simulation currently takes approximately 36 h to complete, limiting the application of current *in-silico* subject-specific technique training methods to high-risk athletic populations. Future research is therefore needed to develop clinically-relevant ACL injury risk estimates to identify high-risk athletes if current subject-specific *in-silico* technique training methods can be effectively utilized.

Generalized kinematic strategies to reduce peak valgus knee loading during sidestepping must be developed for ACL injury risk to be reduced in heterogeneous athletic populations. A generalized kinematic solution would make it possible for coaches and/or clinicians to train athletes to sidestep with reduced valgus knee loading. *In-silico* patient-specific gait modifications have been successfully used to re-train a high functioning osteoarthritis (OA) patient to walk with reduced peak adduction (varus) knee moments and OA related knee pain (Fregly et al., 2007). "Medial-thrust gait", which in general terms focuses on increasing support limb flexion and decreasing the size of the moment arm between the knee joint center and GRF vector during stance was the generalized kinematic strategy identified by Fregly et al. (2007). "Medial-thrust gait" training has since been proven effective in reducing peak varus knee moments in both a single healthy male (Schache et al., 2008) and elderly male OA patient (Walter et al., 2010).

From the nine critical joint coordinates used by each simulation to reduce valgus knee moments, two generalized kinematic strategies were identified. One strategy involved increasing stance foot plantar flexion, while the second was to re-direct the WB CoM medially, towards the desired change of direction pathway.

The ankle plantar flexion strategy used by six of the nine simulations likely reduced peak valgus knee moments by changing the position of the ankle joint center relative to the GRF vector during WA. Small changes in joint center position have non-linear effects on proximal joint torques along the kinematic chain (Reinbolt et al., 2007). Changing joint center position also has differing effects on joint torques expressed in the M/L and A/P DoF (Reinbolt et al., 2007). These non-linear relationships make it difficult to identify how plantar flexion influences valgus knee moments during sidestepping. Additionally, without a foot-contact model, it is unlikely that these results would be observed in an experimental setting. We are therefore limited in our ability to make conclusions associated with plantar/dorsi flexion and valgus knee loading, leaving this relationship to be verified with future research.

Re-positioning the CoM medially, towards the desired change of direction pathway is a motor control strategy used during change-of-direction tasks (Patla et al., 1999) and similar to one of three technique recommendations used to reduce peak valgus knee moments during sidestepping (Dempsey et al., 2009); meaning this kinematic strategy can be learned by athletic populations. A secondary benefit of this generalized technique recommendation is that individuals can develop unique motor control strategies to successfully learn this technique modification. This generalized technique modification subsequently represents a form of subject-specific technique training.

One concern for using RRA to reduce peak valgus knee moments during UnSS is that the sidestep motion may not be preserved pre-to-post kinematic optimization. These concerns are addressed in three ways. First, the goal of RRA is to reduce the residual forces and moments held in the pelvis, producing a torque driven simulation that is dynamically consistent with the experimental GRF's measured during UnSS. This is an important consideration, as these external forces are needed to redirect the WB CoM during sidestepping (Jindrich et al., 2006) and are therefore a fundamental component of a realistic simulation of the sidestep motion. Second, the motion of all nine simulation's CoM were directed medially, towards the desired direction of travel, making an UnSS look more like a pre-planned sidestep (Houck et al., 2006). Third, supplementary video data published with this manuscript shows the sidestep motion was indeed maintained pre-to-post kinematic optimization.

Previous literature has shown that *in-silico* technique modifications are effective in reducing peak varus knee moments in multiple case studies (Fregly et al., 2007; Schache et al., 2008; Walter et al., 2010). Findings from this study must now be tested in a controlled laboratory setting, with large heterogeneous athletic populations. Once the efficacy of directing the WB CoM medially during sidestepping to reduce peak valgus knee moments is established, it can then be recommended to heterogeneous athletic populations.

These methods possess an enormous potential within the injury prevention literature. We were capable of identifying a single kinematic solution to reduce valgus knee loading during a complex, multi-body, dynamic movement with an enormous solution space. Nevertheless, we encourage future *in-silico* research to build upon these findings. For example, with a foot contact model, additional kinematic strategies to reduce valgus knee moments during UnSS and ACL injury risk may be identified. Alternate solutions may also be possible if the optimization criterion was amended to both reduce valgus knee loading and optimize sidestep performance. It is through this rigor that additional casual information may become available and assist in the development of short, concise and effective technique training protocols designed to reduce ACL injury risk. It is through this process we can more effectively translate ACL focused research into injury prevention practice for the community level athlete (Finch, 2006).

Conflict of interest

There were no financial or personal relationships with other people or organizations that could have biased the presented work.

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Appendix A. Supplementary information

Supplementary data associated with this article can be found in the online version at 10.1016/j.jbiomech.2012.02.010.

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