Kinematic analysis of functional lower body perturbations

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Abstract:

Background. Sudden changes in direction on a single weight-bearing-limb are commonly associated with injury to the lower extremity. The purposes of this study were to assess the between day reliability of hip, knee, and ankle kinematic displacements achieved with internal and external femur-on-weight-bearing-tibia rotation perturbations and to determine the effect of these perturbations on three dimensional hip, knee and ankle kinematics.

Methods. Twenty recreationally active, healthy college students with no history of significant orthopedic injury (10 male, 10 female) were subjected to a forward and either internal or external rotary perturbation of the trunk and thigh on the weight-bearing-tibia while three dimensional kinematics were simultaneously collected. The protocol was repeated 24–48 h later to assess reliability.

Findings. External perturbations resulted in significant internal rotation (IR) of the tibia on the femur (mean 7.3 (SD 3.9°)) and IR of the femur on the pelvis (mean 6.8 (SD 5.4°)) ($P < 0.05$). Internal perturbations resulted in significant external rotation (ER) of the tibia on the femur (mean 6.8 (5.9°)) and ER of the femur on the pelvis (mean 10.7 (SD 96.1°)) ($P < 0.05$). Additionally the external perturbation results in a significantly greater knee valgus (mean 3.6 (SD 2.2°)) position while the internal perturbation results in a significantly greater knee varus position (mean 2.3 (SD 3.5°)) ($P < 0.05$). External perturbation hip and knee total joint displacements revealed moderate to strong reliability (Intraclass Correlation Coefficient2,k = 0.67–0.94) while internal perturbations revealed slightly higher Intraclass Correlation Coefficients2,k(0.80–0.96).

Interpretation. The lower extremity perturbation device provides a consistent external and internal perturbation of the femur on the weight-bearing-tibia. The observed transverse and frontal plane kinematics are similar to motions observed during cross-over and side-stepping tasks.
Keywords: Knee; Hip; Injury mechanism

Article:

INTRODUCTION

Perturbations have commonly been used as models of injury mechanism to understand the neuromuscular and biomechanical responses to potentially deleterious actions (Da Fonseca et al., 2004; Ferber et al., 2003; Myers et al., 2003; Wojtys et al., 2003). These perturbation models have been applied while walking, sitting, and standing. One model that utilizes a standing, functional perturbation is the lower extremity perturbation device (LEPD) (Shultz et al., 2000).

The LEPD was designed to produce a sudden, forward and either internal or external rotation perturbation of the trunk and femur relative to the weightbearing-tibia. Such perturbations may likely occur during physical activity when an individual moving one direction plants their foot with a single limb and begins to change direction in some manner. Initial work established the reliability and validity of this weight- bearing perturbation model (Shultz et al., 2000), and found neuromuscular responses differ markedly from those previously reported using seated or partial weight-bearing models with the muscles at rest. Subsequent work examined how sex (Shultz et al., 2001), limb alignment (Shultz et al., 2002), foot orthoses (Rose et al., 2002) and knee joint laxity (Shultz et al., 2004) influence neuromuscular control of the knee, to better understand the factors that may contribute to reduced musculoskeletal stability during similar weight-bearing perturbations. Females were found to be more reliant on the quadriceps to stabilize the knee in response to the perturbation than males (Shultz et al., 2001). These findings may be in part be attributed to sex differences in anatomical structure (e.g., knee laxity) (Shultz et al., 2004).

One reported limitation of this work to date is that it has evaluated only neuromuscular response characteristics. The implications of the observed differences in neuromuscular control strategies on biomechanical factors (i.e., joint kinematic and kinetics) are not yet clear. Future studies combining neuromuscular and biomechanical analyses will enhance this model's ability to assess the impact of injury risk factors on functional joint and postural stability. Thus, the purposes of this study were to assess the between day reliability of hip, knee, and ankle kinematic displacements achieved with internal and external femur-on-weight-bearing-tibia rotation perturbations and to determine the effect of these perturbations on three dimensional hip, knee and ankle kinematics. We chose not to include hypotheses for the reliability data as it would be expected that if the reliability data were not good, we would not be able to reliably measure total joint displacements. It was hypothesized that rotational patterns at the hip and knee would consistently be opposite between the two perturbation types with external rotation (ER) perturbations resulting in internal rotation (IR) of the hip and knee and IR perturbations resulting in ER of the hip and knee.

METHODS

Subjects

Twenty (mean (SD):10 male: 26.5 (4.2) years, 177.2 (5.8) cm, 84.4 (17.6) kg and 10 female: 23.2 (3.6) years, 165.8 (4.6) cm, 62.9 (7.6) kg) recreationally active athletes participated in this study. Recreational activity was defined as engagement in exercise 3 times per week for 30 min. in duration. Upon arrival to the laboratory, each subject read and signed a written informed consent approved by the University Institutional Review Board.

Instrumentation

Kinematic data for the head, thorax, pelvis, thighs, shanks, and dominant foot were collected at 140 Hz using an electromagnetic tracking system (Ascension Star Hardware, Ascension

Technology, Burlington, VT, USA) (Motion Monitor Software, Innovative Sports Training, Chicago, IL, USA). A Bertec Force Plate, (Type 4060 non-conducting Bertec Corporation, Columbus, OH, USA) acquired ground reaction forces at 1000 Hz. A lower extremity perturbation device was used to produce an external or internal rotational perturbation of the femur relative to the tibia in single-leg stance while positioned with the weight-bearing-knee in 30° of flexion. The design and ability of this device in obtaining neuromuscular response data have been previously reported (Shultz et al., 2000).

Procedures

After each subject read and signed the informed consent form, they were prepared for kinematic data collection. Kinematic setup involved the examiner attaching six-degree-of-freedom position sensors (Ascension Technologies, Burlington, VT, USA) to each of the following

Fig. 1. Initial positioning — kevlar release cables attached to belt via load cells.

anatomical sites: occiput of skull, spinous process of C7, sacrum, lateral aspect of mid-shaft of both femurs, medial aspect of the mid-shaft of both tibias, and the mid- shaft of third metatarsal of the dominant foot. Sensors on the thorax and skull were secured with neoprene wraps while all others were secured directly to the skin via double-sided tape and covered with athletic pre-wrap and tape.

The subject's leg dominance was determined by asking with which foot the subject preferred to kick a soccer ball. An electrogoniometer (Penny and Giles, Santa Monica, CA, USA) was placed on the medial aspect of the dominant knee to visually monitor joint flexion angle prior to each trial. Kevlar cables were attached to the lateral aspects of the subject's hips via a waist harness, positioned over the subject's anterior superior iliac spines, and to wall mounted release mechanisms that sent analog signals to the data acquisition computer. The cables to the release mechanisms were height adjusted at the wall so that the cables were parallel to the floor when the subject was in the testing position.

Each subject was instructed to stand on the dominant foot in the middle of the force plate in an upright man ner while passively leaning into the cables with arms folded across the chest. An examiner then instructed the subject to bend his/her knee to 30° as indicated by the electrogoniometer. A plumb bob was used by the examiner to ensure that the greater trochanter of the dominant leg was positioned directly superior to the area between the 1st metatarsophalangeal joint and the navicular bone. At this point the subject adjusted his/her center of pressure (CoP) in the anterior–posterior plane with a feedback monitor positioned in front of the subject. The CoP feedback monitor was designed to ensure that the subject's CoP was posterior to the 1st metatarsophalangeal joint and anterior to the navicular bone. To ensure that equal cable forces were placed on the hips, an examiner observed light emitting diode monitors displaying forces from the transducers (WMC1000, Interface, Scottsdale, AZ, USA) imbedded in each of the cables.

Once proper position was achieved, one cable was released producing either an ER (Fig. 2) or IR (Fig. 3) perturbation of the femur on the weight-bearing-tibia. All subjects were instructed to relax their hips into the cables and then react to the perturbation by attempting

Fig.2. External Perturbation

Fig.3. Internal Perturbation

to maintain balance. Practice trials were performed until the subject achieved appropriate positioning and became comfortable with the task. Twenty trials (10 ER, 10 IR) were acquired in a randomized order to minimize anticipatory responses. All procedures were repeated exactly as described on a second day within 24–72 h of the first data collection.

Data reduction

Kinematic data for the hip, knee and ankle were low passed at 12 Hz using a 4th order, zero lag digital Butterworth filter. A segmental reference system was defined for all body segments with the positive Z-axis defined as the medial to lateral axis; the positive Y-axis defined as the distal to proximal longitudinal axis; and the positive X-axis defined as the posterior to anterior axis. Three-dimensional hip, knee, and ankle flexion angles were calculated using Euler angle definitions with a rotational sequence of Z Y' X''. Increasing amounts of hip flexion, hip internal rotation, hip adduction, knee extension, knee internal rotation, knee adduction, and ankle flexion were defined as positive.

Three-dimensional total joint displacements of the hip, knee and ankle were defined as the difference between the initial joint angle at the time of cable release, and the joint angle at the time the body's center of mass velocity changed to a negative slope (representing contralateral foot contact). The center of mass model used for this calculation accounted for 88.5% of the entire mass of the body (Zatsiorsky et al., 1990). Both upper extremities and the non-stance foot were excluded due to instrumentation limitations.

Statistical analyses

All dependent variables were calculated for each of the 10 trials for each of the internal and external conditions. As this study was part of a larger project in which surface electromyography (EMG) data were acquired, we used previously reported EMG criteria for trial selection, averaging the first five acceptable trials for statistical analyses (Shultz et al., 2000). Intraclass correlation coefficients (ICC 2,k) and standard errors of measurement (SEM) were computed to assess day-to-day reliability of the hip, knee, and ankle kinematic variables of total joint displacements as well as initial joint angles. Repeated measure ANOVAs compared the total joint displacements of the two perturbations. Statistical significance was set a priori at $P < 0.05$.

RESULTS

Means, standard deviations and reliability coefficients for initial and peak joint displacements, for external and

Table 1
Reliability of external perturbation initial joint angles

	$ICC_{2,k}$(SEM°)	Day 1 mean (SD)°	Day 2 mean (SD)°
Initial joint angle			
Hip flex	0.53 (5.1)	17.1 (6.6)	20.5 (7.4)
Hip rot	0.87 (3.0)	5.9 (6.3)	5.2 (8.4)
Hip add	0.76 (3.8)	9.1 (6.2)	6.7 (7.9)
Knee flex	0.41 (3.5)	−30.5 (4.4)	−30.5 (4.6)
Knee rot	0.88 (2.0)	−0.4 (5.7)	0.4 (5.2)
Knee add	0.91 (1.5)	0.8 (4.8)	0.5 (4.0)
Ankle flex	0.82 (3.0)	95.0 (7.1)	96.4 (5.2)

internal perturbations respectively, are listed in Tables l–4. External and internal perturbations revealed mostly moderate to strong measurement consistency in initial joint angles (ICCs = 0.70–0.91), hip flexion (0.53 and 0.66) and knee flexion (0.41 and 0.51) initial joint angle ICCs were consistently lower (Tables 1 and 3). External perturbation hip and knee total joint displacement ICCs revealed moderate to strong reliability (0.67–0.94)

Table 2
Reliability of external perturbation total joint displacements

	$ICC_{2,k}$(SEM°)	Day 1 mean (SD)°	Day 2 mean (SD)°
Total joint displ.			
Hip flex	0.84 (2.0)	−6.0 (4.3)	−6.8 (4.9)
Hip rot	0.94 (1.4)	6.8 (5.4)	6.7 (5.2)
Hip add	0.67 (2.0)	−3.3 (2.5)	−3.2 (3.4)
Knee flex	0.94 (1.4)	2.3 (4.1)	1.7 (5.8)
Knee rot	0.81 (1.7)	7.3 (3.9)	6.5 (3.6)
Knee add	0.85 (0.8)	−3.6 (2.2)	−3.1 (1.8)
Ankle flex	0.61 (1.6)	−1.5 (2.1)	−1.9 (2.5)

Table 3
Reliability of internal perturbation initial joint angles

	$ICC_{2,k}$(SEM°)	Day 1 mean (SD)°	Day 2 mean (SD)°
Initial joint angle			
Hip flex	0.66 (4.1)	17.8 (6.8)	20.3 (7.0)
Hip rot	0.82 (3.6)	4.5 (6.5)	3.6 (8.5)
Hip add	0.70 (4.4)	9.1 (6.4)	6.7 (8.0)
Knee flex	0.51 (2.9)	−31.3 (3.9)	−30.5 (4.2)
Knee rot	0.87 (2.1)	−0.9 (5.9)	0.0 (5.3)
Knee add	0.90 (1.5)	1.0 (4.9)	0.9 (4.4)
Ankle flex	0.75 (3.9)	95.8 (7.1)	94.8 (7.8)

Table 4
Reliability of internal perturbation total joint displacements

	$ICC_{2,k}$(SEM°)	Day 1 mean (SD)°	Day 2 mean (SD)°
Total joint displ.			
Hip flex	0.85 (1.9)	−4.9 (4.8)	−5.4 (4.8)
Hip rot	0.86 (2.6)	−10.7 (6.1)	−11.7 (6.9)
Hip add	0.80 (1.5)	−1.4 (2.9)	−1.2 (3.3)
Knee flex	0.96 (1.4)	2.9 (6.4)	2.2 (7.2)
Knee rot	0.94 (1.5)	−6.8 (5.9)	−7.6 (5.9)
Knee add	0.93 (1.0)	2.3 (3.5)	2.8 (3.3)
Ankle flex	0.58 (3.9)	1.3 (3.0)	2.5 (6.0)

Table 5
Total joint displacements for external and internal perturbations

	External mean (SD)°	Internal mean (SD)°	*P* value
Total joint displ.			
Hip flex	−6.0 (4.3)	−4.9 (4.8)	$F_{(1,18)} = 1.53$; $P = 0.232$
Hip rot*	6.8 (5.4)	−10.7 (6.1)	$F_{(1,18)} = 61.07$; $P \leqslant 0.001$
Hip add	−3.3 (2.5)	−1.4 (2.9)	$F_{(1,18)} = 4.09$; $P = 0.058$
Knee flex	2.3 (4.1)	2.9 (6.4)	$F_{(1,18)} = 0.40$; $P = 0.536$
Knee rot*	7.3 (3.9)	−6.8 (5.9)	$F_{(1,18)} = 48.35$; $P \leqslant 0.001$
Knee add*	−3.6 (2.2)	2.3 (3.5)	$F_{(1,18)} = 31.39$; $P \leqslant 0.001$
Ankle flex*	−1.5 (2.1)	1.3 (3.0)	$F_{(1,18)} = 27.91$; $P \leqslant 0.001$

* Significant difference between perturbation condition.

(Table 2) while internal perturbations revealed slightly higher ICCs at the hip and knee (0.80–96) (Table 4). Ankle flexion ICC values for both perturbations were somewhat lower in magnitude (0.61 and 0.58, respectively).

Average initial position (Fig. 1) was characterized by 17–20° of hip flexion, 3–6° of hip internal rotation, 6–9° of hip adduction, 30–31° knee flexion, less than 1 ° of knee rotation, less than 1 ° of knee ab/adduction, and 4–6° of ankle dorsiflexion. The external and internal perturbation comparisons of total joint displacements on day 1 revealed significant differences in hip rotation ($F_{(1,18)} = 61.07$, $P \leq 0.001$), knee rotation ($F_{(1,18)} = 48.35$, $P \leq 0.001$), knee adduction ($F_{(1,18)} = 31.39$, $P \leq 0.001$), and ankle flexion ($F_{(1,18)} = 27.91$, $P \leq 0.001$). (Table 5). The ER perturbations resulted in IR of the femur on the pelvis (hip) and IR of the tibia on the femur (knee) while IR perturbations resulted in ER of the femur on the pelvis (hip) and ER of the tibia on the femur (knee). Additionally, the knee assumed a valgus position as a result of ER perturbations, while IR perturbations resulted in a varus knee position (Table 5). Although there was a strong statistical trend for hip adduction between perturbation condition ($P = 0.058$), with the ER perturbation resulting in ~2° more hip abduction, we feel that this likely is functionally or clinically non-significant as the subjects started in 6–9° of hip adduction thus maintaining a hip adducted position as a result of either perturbation. Both perturbations resulted in 5– 6° of hip extension and l–3° of knee extension. Representative single trial hip and knee time-series kinematic curves of external perturbation and internal perturbation are found in Figs. 4 and 5, respectively.

DISCUSSION

The primary findings of the study were that consistent joint displacements can be obtained from the LEPD on repeated test days. Additionally, the hypotheses of ER perturbations resulting in IR of the hip and knee and IR perturbations resulting in ER of the hip and knee were accepted.

Reliability

Initial Joint Angle ICCs were calculated to ensure consistent pre-perturbation positioning and resulted in mostly moderate to strong consistency (with the exception of knee and hip flexion). The low knee flexion ICCs were not surprising, given the small amount of variance between subjects given the standardized position of 30° knee flexion. Detailed examination of the hip data did not reveal a small amount of hip flexion variance between subjects. We speculate that this lower ICC may

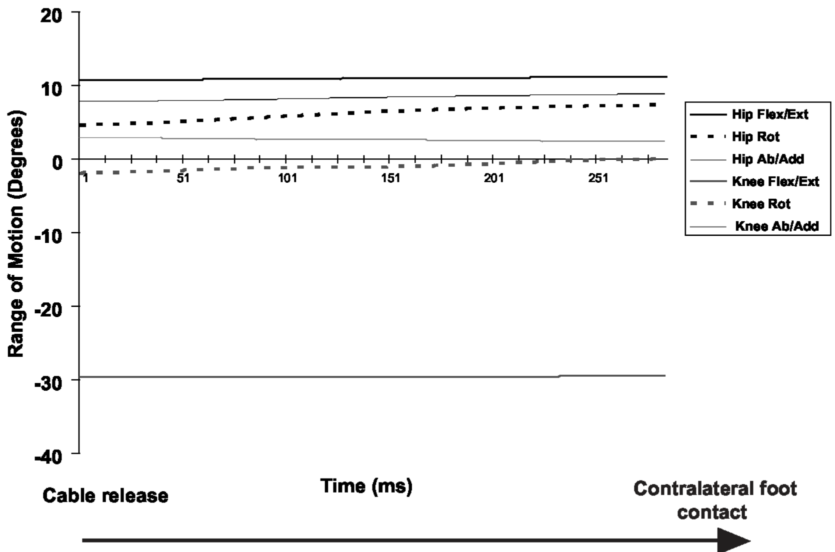

Fig. 4. External perturbation representative hip and knee kinematics from cable release to bilateral foot contact. Increasing amounts of hip flexion, hip internal rotation, hip adduction, knee extension, knee internal rotation, and knee adduction were defined as positive.

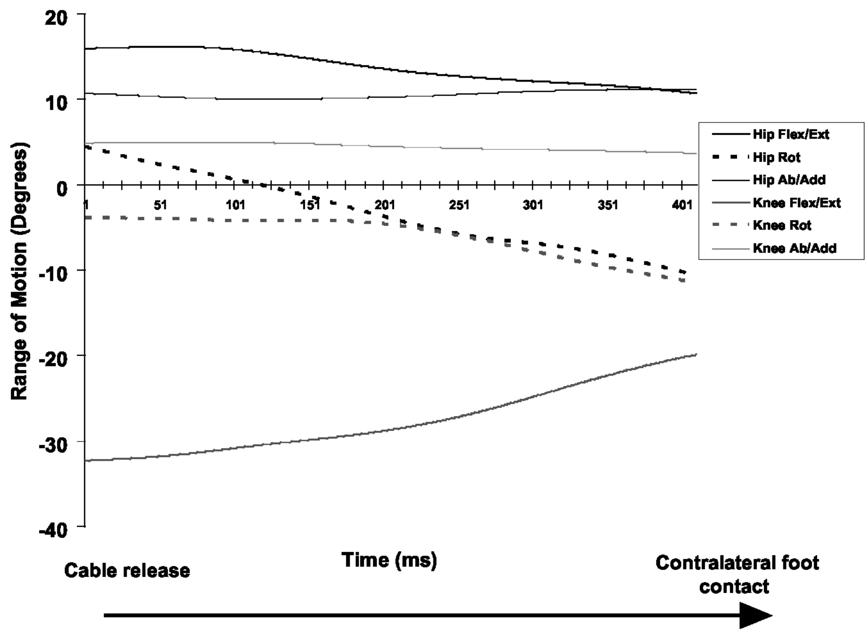

Fig. 5. Internal perturbation representative hip and knee kinematics from cable release to bilateral foot contact. Increasing amounts of hip flexion, hip internal rotation, hip adduction, knee extension, knee internal rotation and knee adduction were defined as positive.

be due to the positioning instructions of “passively leaning” into the waist harness to optimally place the CoP. It is likely that the subjects did this through a less reliable method of slightly adjusting both the hip and trunk flexion angles.

Total joint displacement ICCs were calculated to assess the between day reliability of the LEPD in producing kinematic displacements. These values were generally moderate to high with the exception of ankle flexion. Again this was attributed to the relatively small amount of variance between subjects for this variable.

Previous work has reported the ability of the LEPD to provide consistent perturbations through maximum postural sway distance measures (Shultz et al., 2000). Day to day postural sway consistency responses following perturbations were reported to range from 0.81 to 0.84 for internal perturbations and 0.69–0.71 for external perturbations (Shultz et al., 2000). These values are somewhat representative of the range of the total joint displacement ICCs measured for external (0.61–0.94) and internal (0.58–0.96) perturbations.

Comparison of joint displacements with external and internal perturbations

Knee rotation

The IR of the knee joint (tibial rotation with respect to the femur) during external perturbations is similar to that observed during a crossover cut maneuver (Houck and Yack, 2003). During a crossover cut the contralateral limb is placed over the stance limb to help redirect the body's center of mass (Besier et al., 2001; Houck and Yack, 2003; Nyland et al., 1999). The IR at the knee during a crossover cut has been associated with the difficulty of anterior cruciate ligament (ACL) deficient subjects completing the maneuver (Houck and Yack, 2001). Healthy subjects during a walking and crossover cutting task produced 8.3 (2.5)° of IR at the knee (Houck and Yack, 2003), which are very similar to our observed range of 7.3 (3.9)° during external perturbations (Table 5). A previous report has sought to determine the effect of medially directed torsional stresses on the knee joint (Wojtys et al., 2003). The device was designed such that subjects activated their muscles while seated in 30° knee flexion. Values of 6.0 (2.2)° (males) and 7.6 (2.8)° (females) of internal knee rotation were reported (Wojtys et al., 2003). These findings suggest that the perturbation created by the LEPD results in reliable knee rotations that are consistent with previous clinical and laboratory studies.

Internal rotation at the knee joint is also associated with the common clinical test of the pivot shift. The pivot shift test combines internal and valgus loading through a range of knee extension angles to assess integrity of the ACL (Kanamori et al., 2000). The external perturbation produced the kinematic pattern associated with the pivot shift (internal knee rotation and knee abduction). The magnitude of the IR displacement as a result of displacement is somewhat less than cadaveric testing reports (22.3 (3.5°)) of the pivot shift maneuver (Kanamori et al., 2000). It would be expected that the cadaveric testing technique would allow for greater knee rotation as it has been demonstrated in a previous perturbation model that the amount of rotation occurring at the tibiofemoral joint is significantly less when the muscle are in an active state (Wojtys et al., 2003). Additionally the increased IR was accompanied by valgus excursion at the knee which together have been often observed as an ACL injury mechanism (Boden et al., 2000; DeMorat et al., 2004; Kirkendall and Garrett, 2000). This suggests that the LEPD creates a realistic perturbation in a weight-bearing model from which to assess the affects of various risk factors on neuromuscular and biomechanical responses.

As the original purpose of the LEPD was to deliver a perturbation that would result in joint motions that are inherent to ACL injury (Shultz et al., 2000), it is important to understand the extent to which the perturbation may actually strain the ACL. As it is non-practical to implant strain transducers in the ACL with the current set-up, comparisons can be made with motions in which the ACL acts as a restraint of motions known to strain the ACL. Cadaveric testing has demonstrated that removing the ACL leads to increases in tibial rotation at 10° and 30° of knee flexion (Andersen and DyhrePoulsen, 1997). However because of instrumentation it was not reported in which direction the increase occurred (internal or external). Others using cadaveric testing methods have suggested that between 60% and 85% of the total increase in tibial rotation after removing the ACL is internal in direction (Lane et al., 1994; Nielsen et al., 1984). Research of in vivo strain testing of the human ACL has reported that over a range of ER torques and low IR torques, strains produced in the ACL during weight-bearing were greater than during non-weight-bearing (Fleming et al., 2001). Thus we feel that there is evidence to suggest that the LEPD provides a perturbation that may in part strain the ACL.

The internal perturbations produced a movement characteristic of the side step maneuver where there must be a component of knee external rotation (femur internally rotating on the stationary tibia) (McLean et al., 1999). During the side-step maneuver the swing leg moves laterally to direct the body's center of mass away from the stance limb (Colby et al., 2000; Cross et al., 1989; Malinzak et al., 2001). McLean et al. (1999) graphically reported less than 10° of external tibial rotation during the stance phase of a side-step maneuver when the knee flexion ranged from — 18° to 35°. These values are consistent with the measures attained in the current study.

Hip rotation

Currently, there is a dearth of research that addresses transverse plane motion of the hip in perturbation models designed to mimic lower extremity injury mechanisms. The internal perturbations do produce the external hip rotation described in the dangerous “position of no return” (Ireland, 1999). This position describes the kinematic characteristics of putting the knee at an increased risk of injury as seen in sport. However, not only has external hip rotation been associated with knee injury, but loss of hip and pelvis control has also been suggested to be a contributing factor (Ireland, 1999). Although, we have been unable to find a report of hip kinematics during the crossover cut, we would expect to see the internal hip rotation observed in our external perturbations. It has been suggested that future work consider the relationship of trunk position to the limb to better understand how the limb transitions the body's center of mass in a different direction (Houck and Yack, 2003; Besier et al., 2001). We feel that the current study is a rudimentary beginning to best understand how the proximal segments interact with a rotational perturbation at the knee.

CONCLUSION

The findings of the current study demonstrate the successful integration of kinematic analyses with this perturbation model, and provide further confirmation that the LEPD provides a consistent external and internal perturbation of the femur on the weight-bearingtibia. The ability of the LEPD to produce statistically different hip and knee rotation patterns in a single leg, weight-bearing position with ER and IR perturbations allows examination of joint motions that are commonly associated with ACL injury mechanisms. With an external perturbation, neuromuscular and biomechanical responses to combined IR of the tibia on the femur, knee valgus, and IR of the femur on the pelvis can be examined. Conversely, biomechanical responses to combined ER of the tibia on the femur, knee varus, and ER of the femur on the pelvis can be examined with the IR perturbation. Future studies combining neuromuscular and biomechanical analyses (including the inter-segmental forces and moments) will help determine the impact that alterations in neuromuscular control actually have on knee joint motion and forces.

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