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Coupling motion between rearfoot and hip and knee joints during walking and single-leg landing

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ABSTRACT

The objective of the current study was to investigate the kinematic relationships between the rearfoot and hip/knee joint during walking and single-leg landing. Kinematics of the rearfoot relative to the shank, knee and hip joints during walking and single-leg landing were analyzed in 22 healthy university students. Kinematic relationships between two types of angular data were assessed by zero-lag cross-correlation coefficients and coupling angles, and were compared between joints and between tasks. During walking, rearfoot eversion/inversion and external/internal rotation were strongly correlated with hip adduction/abduction ($R = 0.69$ and $R = 0.84$), whereas correlations with knee kinematics were not strong ($R \leq 0.51$) and varied between subjects. The correlations with hip adduction/abduction were stronger than those with knee kinematics ($P < 0.001$). Most coefficients during single-leg landing were strong ($R \geq 0.70$), and greater than those during walking ($P < 0.001$). Coupling angles indicated that hip motion relative to rearfoot motion was greater than knee motion relative to rearfoot motion during both tasks ($P < 0.001$). Interventions to control rearfoot kinematics may affect hip kinematics during dynamic tasks. The coupling motion between the rearfoot and hip/knee joints, especially in the knee, should be considered individually.

1. Introduction

The kinematics of the foot and ankle affect proximal joints kinematics, such as hip and knee joints, during both static and dynamic conditions (Khamis and Yizhar, 2007; Resende et al., 2015; Tateuchi et al., 2011). This linkage between foot/ankle and the proximal joints may contribute to musculoskeletal injuries in the lower limbs (Chuter and Janse de Jonge, 2012). For example, the pathology of patellofemoral pain syndrome (Barton et al., 2009) and medial tibial stress syndrome (Viitasalo and Kvist, 1983) are reported to be related to dynamic foot function. In addition, knee valgus, which is a risk factor for anterior cruciate ligament injury, has been partially attributed to excessive foot pronation (Joseph et al., 2008). Excessive foot and ankle motion may be associated with a variety of sports injuries in the lower limbs.

The effects of foot and ankle kinematics on lower limb joint kinematics have been investigated in a small number of studies. Induced hyperpronation of the foot by wedges was found to result in increases in internal rotation of both the knee joint and the hip joint during standing (Khamis and Yizhar, 2007), increased hip internal rotation during single-leg standing (Tateuchi et al., 2011), and increased internal rotation of the hip joint, femur and shank, as well as changes in the temporal pattern of knee internal rotation during walking (Resende et al., 2015). However, these studies examined the effect of the hyperpronation of the foot induced by wedges, which may be beyond the range of normal foot motion. In a previous study that did not induce foot motion, rearfoot eversion was found to be synchronized with hip internal rotation (Souza et al., 2010) and correlated with hip adduction and shank internal rotation during the stance phase of walking (Barton et al., 2012). However, to our knowledge, the effects of rearfoot kinematics on the kinematics of the hip and knee joints have only been examined during walking, and have not been examined during sports-related tasks such as jump-landing, which is involved in a variety of sports and is associated with musculoskeletal injuries of the lower limbs (Doherty et al., 2016; van der Does et al. 2016). Thus, examining coupling motion during a landing task could provide basic information

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for assessment of joint kinematics and for prevention and rehabilitation of musculoskeletal injuries of the lower limbs in clinical settings.

Lafortune et al. (1994) examined the effects of a 10° pronation wedge and a 10° supination wedge on knee kinematics using bone-pins during walking. Their results revealed only minor changes in the knee angular pattern, suggesting that foot kinematics had a weak effect on knee joint kinematics during walking. Although the findings suggested that tibial rotation induced by pronation/supination wedges was resolved at the hip joint rather than the knee joint, hip joint kinematics were not measured in the previous study (Lafortune et al., 1994). Importantly, it is currently unclear whether foot kinematics have a stronger association with the hip or knee joint during dynamic tasks. It is important for clinicians to understand the interrelationships within the lower limb kinematics to address malposition of the hip or knee joint. The current study had three main aims: to investigate the kinematic relationships between the rearfoot and the hip and knee joints during walking and single-leg landing, to investigate whether the relationship between the rearfoot and hip joint differed from those between the rearfoot and knee joint, and to compare those relationships between walking and single-leg landing. We hypothesized that rearfoot kinematics would be associated with hip joint kinematics, whereas rearfoot kinematics would not be associated with knee joint kinematics during both tasks.

2. Methods

2.1. Subjects

Twenty-two healthy university students participated in this study (11 males, 11 females, age: 21.9 (1.1) years old, height: 167.2 (8.4) cm, body weight: 57.4 (6.6) kg). A priori power analysis in G*Power 3.1.7 was performed using the correlation coefficients between rearfoot and hip joint motion in a previous study (Souza et al., 2010). As a result, at least 22 subjects were required to achieve statistical power of 80% with an alpha level of 0.05 for the correlation analyses. All participants had no history of surgery or fracture in the lower limbs, and had no musculoskeletal injuries within the past 6 months. Because the dominant side (the side used for kicking a ball) was the right leg in all subjects, the right lower limbs were tested and analyzed. The experiments were performed after gaining ethical committee approval from the University Institutional Review Board. Informed consent was obtained from all subjects.

2.2. Procedure

Six high-speed digital cameras (Hawk cameras, Motion Analysis Corporation, Santa Rosa, CA, USA) and a force plate (Type 9286, Kistler AG, Winterthur, Switzerland) were time-synchronized and used for motion analysis during walking and single-leg landing. Reflective markers were attached to the bilateral anterior superior iliac spine, sacral, lateral thigh, and lateral and medial femoral epicondyles. Markers of the shank and foot were attached to the tibial tuberosity, the head of the fibula, lateral and medial malleoli, Achilles' tendon attachment, posterior surface of the calcaneus, peroneal tubercle, sustentaculum tali, tuberosity of the navicular, base of the first, second and fifth metatarsal, head of the first, second and fifth metatarsal, and head of the proximal phalanx of the hallux, based on the Rizzoli multi-segment foot model (Fig. 1) (Leardini et al., 2007). EvaRT 4.3.57 (Motion Analysis Corporation) software was used to record the marker coordinates during each task, sampled at 200 Hz for kinematic data and 1000 Hz for force data.

For the walking task, subjects walked at their natural speed. For single-leg landing, subjects dropped from a 30-cm box from their left leg, and landed with the right leg on the force plate. Subjects practiced up to 10 trials of each task before recording, and performed three successful trials for each task. Trials in which the entire right foot landing on the force plate, the left foot did not touch the force plate, and the subject did not lose balance during testing were defined as successful trials.

Fig. 1. Marker location.

2.3. Data collection and reduction

Kinematic data were low-pass filtered using a 4th order Butterworth filter with a 6 Hz cutoff frequency. Hip and knee joint angles were calculated using the traditional lower limb model (Helen Hayes model), and the rearfoot angle was calculated using the Rizzoli multi-segment foot model (Leardini et al., 2007) using Visual 3D software (C-Motion Inc., Germantown, MD, USA). The Rizzoli multi-segment foot model has five segments, as follows: shank, rearfoot, midfoot, forefoot and hallux. In the current study, the rearfoot angle with respect to the shank was calculated according to the joint coordinate system (Grood and Suntay,