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Increased double limb support times during walking in right limb dominant healthy older adults with low bone density

Paul S. Sung

Department of Physical Therapy, Motion Analysis Center, Herbert H. and Grace A. Dow College of Health Professions, Central Michigan University, Health Professions Building 1220, Mt. Pleasant, MI, 48859, United States

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ABSTRACT

Background: Older adults with low bone mineral density (BMD) may exhibit early markers for physical frailty; however, there is a lack of understanding of the dominant limb support patterns during walking.

Research question: The purpose of this study was to investigate limb support times during walking in healthy older adults with low BMD.

Methods: Seventy-seven right limb dominant older adults (48 female subjects, 29 male subjects) participated in the study. Each participant's BMD (g/cm^2) was measured by dual-energy x-ray absorptiometry (DEXA), and a motion capture system was utilized to measure temporal-spatial gait parameters (cadence, speed, stride length, and limb support times). The limb support times included initial double limb support (IDS), single limb support (SS), and terminal double limb support (TDS) in the stance phase.

Results: Those limb support times were significantly different ($F = 44.28$, $p = 0.001$) and demonstrated an interaction with dominance ($F = 9.44$, $p = 0.003$). In stance phase, the IDS was longer on the non-dominant limb ($t = -3.07$, $p = 0.003$); however, the TDS was significantly longer on the dominant limb ($t = 3.07$, $p = 0.003$). The stride length was longer on the dominant limb ($t = 2.45$, $p = 0.02$) and was positively associated with single limb support on the dominant limb ($r = 0.34$ $p = 0.001$) compared with the non-dominant limb ($r = -0.29$, $p = 0.001$).

Significance: This longer stride length and single limb loading pattern on the dominant limb could increase asymmetrical balance in the stance phase. The gait modification strategies used to increase stride length on the non-dominant limb may allow for more functional symmetrical gait. Further investigation of asymmetric limb support patterns and gait modification strategies might be needed to enhance functional gait performance.

1. Introduction

Approximately one in three adults aged 65 years and older experiences a fall each year with approximately one-third resulting in a fracture in the lower limbs [1]. Falls and the related consequences represent a significant safety problem in daily activities, and those who have fallen or who have a gait or balance problem are at a higher risk of future falls [1,2]. However, little is known about limb support patterns and gait parameters in healthy older adults with low bone density (BMD).

A recent study reported that BMD asymmetry may be due to previous changes in the loading pattern during walking that might have led to asymmetric bone deterioration [3]. This adaptive bone response dominates the phenomenon of individualized geometry as can be deducted from the interchanging in density distribution patterns with interchanged loading conditions [4]. These results indicated that inclusion of specific loading conditions influences the stress distribution on the foot, and, hence, influences the correlation between stress distributions from walking and changes in BMD. However, there is a lack of understanding on asymmetry between the lower limbs, which might be disrupted in the cyclic nature of gait patterns.

It has been reported that older adults may be at risk of accelerated loss of BMD from muscle atrophy, asymmetrical gait, and decreased strength of the ankle muscles on the affected-side [5]. These impairments may support the hypothesis that the relationship between loading and mechanical competence is generally preserved for frequent motor activities, such as walking [6]. The role of lower limb symmetry in human motor performance and the role of neuromuscular aspects have been reported [7–9].

Our previous studies reported an asymmetrical gait pattern with a longer double limb support pattern for the possible dominant limb dependency and postural control strategies [10,11]. However, it is still unclear if these differences are related to a specific limb support pattern in healthy older adults with low BMD. It would be important to investigate specific

E-mail address: sung1p@cmich.edu.

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limb support patterns between the lower limbs while considering individual factors such as age and body mass index (BMI).

Several studies reported that BMI and alignment were associated with different joint loading measures; alignment was more closely associated with the asymmetry or imbalance of the knee compartments [12–14]. The duration of each aspect of the stance phase might decrease as walking speed increases, although the lower limbs require one step negotiation to detect alterations of the motor pattern [7,15]. The reduced walking speed and increased double limb support could be indicative of efforts aimed at improving gait stability [16–18].

The gait cycle is a repetitive pattern that depends on two identical sequences of lower limb motion that simultaneously advance the body along the desired line of progression. As the body moves forward, one limb provides support while the other limb advances in preparation for its role as the support limb. The functional demand should be related to the contribution of each limb in carrying out the tasks of propulsion to control asymmetries [8,19]. During double stance periods, the limbs typically do not share the load equally, as asymmetry on the dominant limb contributes disproportionately to support during walking [20].

Our study only included participants who are right lower limb dominant since there is a notion that humans are typically right limb dominant for activities requiring mobilization and use the contralateral limb for activities requiring postural stabilization [8,21,22]. However, the limb support patterns for right limb dominant participants have not been carefully investigated. For example, preferred limb performance can differ from the contralateral limb since the role of lower limb symmetry might focus on the neuromuscular aspects [9].

In order to identify pathological gait patterns, the objective measurement of specific limb support times in the stance phase are essential. Functional bilateral activities, such as gait, are related to the contribution of each limb in carrying out the tasks of propulsion to control asymmetries [8,19]. In our study, the stance phase was subdivided into initial double stance (IDS), single limb stance (SS), and terminal double limb stance (TDS). The purpose of our study was to investigate limb support times during walking in right limb dominant healthy older adults with low bone density. We hypothesized that the IDS time would increase on the dominant limb in the gait cycle.

2. Methods

2.1. Target population

Subjects were recruited from the local community; and the procedures, goals, and potential risks of the study were explained to them. Those subjects who expressed interest in the study became eligible for the study, and the subjects who met study inclusionary criteria participated in the study. The study protocol was approved by the Institutional Review Board, and all participants signed an informed consent form.

Subjects were eligible to participate if they: 1) were 45 years of age or older and right limb dominant, 2) were able to walk without pain referral into the lower extremities, 3) were free from lower limb injury and a history of neurological disorder, and 4) had no risks associated with participation in the study as confirmed by the subjects' family physicians. Subjects were excluded from participation if they: 1) had a diagnosed psychological illness that might interfere with the study protocol and/or 2) had overt neurological signs (sensory deficits or motor paralysis).

The participant's right lower limb was regarded as the dominant side for all subjects since they preferred to use the right limb to kick a ball to a target as well as to step out first when initiating gait [8,21,22].

2.2. Outcome measures

The motion analysis was performed using a computer-aided video system in the motion analysis laboratory. Before final inclusion, all subjects underwent a functional assessment of the lower extremities by a physical therapist. It was ensured that the subjects met the inclusionary and exclusionary criteria based on the clinical assessment in order to minimize variations of those measurements. For example, standard tape measurements were used to determine the leg length discrepancy (LLD) of each subject in the supine position [23]. The measurements were completed 3 times from the anterior superior iliac spine (ASIS) to the distal tip of the medial malleolus with the nearest 5 mm reading, and the examiner erased the previous mark each time.

Subjects wore comfortable clothes and were instructed to walk barefoot at a self-selected speed on a 10 m walkway. Gait kinematics and kinetics were recorded with equipment from the Motion Analysis System (Motion Analysis Corporation, Santa Rosa, CA) with six infrared cameras capturing three-dimensional full body kinematic motions sampling at 120 Hz.

Kinematic data from three trials for each subject were exported and normalized to 0–100% of a gait cycle. The data were collected from the unloaded platform to determine the zero offset. All kinematic data were filtered using a fourth-order, zero-lag, Butterworth low-pass-filter with a 6 Hz cutoff frequency. Digital video and force plate data were then imported using OrthoTrack 6.5 software (Motion Analysis Corporation, Santa Rosa, CA, USA). The Cortex software was utilized for handling data calibration, tracking, and post- processing for the results of temporal spatial parameters.

The modified Helen Hayes full trunk (with head) reflective marker set was attached to each subject with adhesive tape rings [24–26] The reflective markers were placed in specified locations on the lower extremities, pelvis, and trunk for the tracking of segment coordinate systems during walking trials. These markers were used to define and track a custom six-degree-of-freedom model.

Two force platforms (AMTI OR6-7, Advanced Mechanical Technology, Inc., Watertown, USA) were used to record the ground reaction forces and the force moments in orthogonal directions at a sampling frequency of 1000 Hz. The stance phase intervals that were analyzed were from 0 to 20%, from 21 to 80%, and from 81 to 100%, representing the IDS, SS, and TDS phases of the gait cycle. The percent of gait cycle was analyzed in limb support times between the dominant and non-dominant limbs.

For the definition of the support time measure, the IDS, SS, and TDS were normalized to stride time. Single support occurs when only one foot is in contact with the ground, and the single support time is the time that has elapsed from the last contact of the opposite footfall to the initial contact of the next footfall of the same foot. Double support occurs when both feet are in contact with the ground simultaneously, and the double support time is the sum of the times elapsed during two periods of double support in the gait cycle. Total support time occurs in the stance phase and includes the IDS, SS, and TDS for each limb. Fig. 1 shows an example of asymmetrical limb support differences based on limb dominance with percent of gait cycle (x-axis) and percent body weight of vertical ground reaction force (y-axis).

The BMD (g/cm^2) of the right heel bone was measured by dual-energy x-ray absorptiometry (DEXA), and a standardized T-score was used to define the group categories [27,28]. Our preliminary results of the reliability test indicated that the intra class correlations ($ICC_{2,1}$) were analyzed based on Cronbach's alpha. The test–retest reliability values ranged from 0.88 to 0.95, which were interpreted as excellent. The reliability of the radiographic measurements ($ICC_{2,1}$ of 0.80–0.97) and the standard error of measurement of 0.3–1.7 mm were consistent [29]. It has been reported that the load transmission characteristics of the joints in the foot have estimated that the talonavicular, as well as the navicular–cuneiform joints, endure the highest stress compared with other joints in the foot [30].

2.3. Statistical analysis

Statistical analyses were completed using IBM Statistics 22