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ABSTRACT

Background: Understanding the biomechanics of the medial longitudinal arch (MLA) may provide insights into injury risk and prevention, as well as function of the arch-supporting structures. Our understanding of MLA deformation is currently limited to sit-to-stand, walking, and running. Material and Methods: Three-dimensional deformation of the MLA of the right foot was characterized in seventeen healthy participants during several simulated activities of daily living. MLA deformation was quantified by both changes in arch length and navicular displacement during the stance phase of three motions: walking, stair ascent, and stair descent. Three levels of load were also considered: no load, a front load (13.6 kg), and a backpack load (13.6 kg). Force platforms and an eight-camera motion capture system were used to collect relevant lower extremity kinetic and kinematic data. Results: Motion type had a significant ($p < 0.05$) effect on navicular displacement and arch length elongation with navicular displacement being greatest during stair descent, while the walking and stair descent conditions showed the greatest increase in arch length. External load did not significantly affect either of these two measures ($p > 0.05$). Conclusion: Differences in the MLA deformation variables resulting from varied dynamic activities of daily living can be greater than those during walking and should be considered. Clinical Relevance: Detailing the mechanics of the MLA may aid in further understanding injuries associated with the MLA, and the results of the current study indicate that these mechanics change based on activity.

Keywords: Medial Longitudinal Arch; Stair Ambulation; Load Carrying;

INTRODUCTION

The medial longitudinal arch (MLA) is a dynamic structure in the foot that deforms in response to loading. Clinically, MLA deformation has been quantified several ways, including navicular drop, a measure that Brody[5] described as the vertical change in height of the navicular tuberosity as a person moves from a seated, subtalar neutral position to a standing position. During sit-to-stand, reported navicular drop values range from 5 to 11 mm[2,5,10,13,29] as the load experienced by each foot increases from about 10% body weight while seated[9] to 50% body weight while standing. When moving from sitting to single-leg stance, navicular drop was 4.5 mm greater than dual-leg sit-to-stand navicular drop,[4] indicating that MLA deformation increases with increasing load. Additional lowering of the arch with greater load is also supported by a cadaveric study.[15] One limitation of the navicular drop test is that it only measures vertical changes in height, while many researchers have reported mediolateral motion of the navicular during gait of the same or greater magnitude.[4,7,31]

In addition to the previously cited studies that documented MLA changes during sit-to-stand, a number of studies have explored MLA kinematics during walking. Walking navicular drop varied greatly between individuals in one study, ranging from 1.7 mm to 13.4 mm[26] compared to seated navicular height. Using three-dimensional analysis to account for mediolateral navicular movement during walking, navicular movement was 7.9 ± 2.5 mm compared to standing navicular height.[7]

As navicular height decreases, arch length typically increases.[6,12] Arch length, defined as the distance between the first metatarsal head and posterior calcaneus, increased by approximately 2% during walking as compared to seated arch length.[6] Gefen[12] used fluoroscopy to evaluate changes in plantar fascia length of two female participants during walking and found the plantar fascia elongated by 9 to 12% compared to its length at early stance phase (before forefoot contact).

Another relevant characteristic of arch kinematics is the timing of when peak deformations occur. Peak changes in arch height/navicular movement during walking occur at 50% to 85%[6,7,16] of stance phase while maximum arch lengthening appears to peak earlier at about 20% of stance and remains relatively constant until 80% of stance.[6] The occurrence of the two ground reaction force peaks at about 20% and 78% of stance phase during walking[16] seem to correspond to peaks in MLA deformation. For example, Kayano[17] observed the MLA elongating due to increases in vertical ground reaction force until foot flat, after which it shortened but then slightly lengthened again when the triceps surae muscles were activated. Simulated walking in cadaver feet has shown plantar fascia tension increased until late stance and was well correlated to Achilles tendon force ($r = 0.76$).[8] Protopapadaki et al.[28] noted that maximum external ankle moments occurred early in stance (~10%) during stair descent and later in stance (~55%) during stair ascent, which closely mimicked the maximum ground reaction force peaks. Therefore, it is feasible to believe much of the MLA collapse is accounted for by muscle activity and ground reaction forces, although this has not been previously quantified.

There are several plausible biomechanical links between MLA deformation and foot/lower extremity pathologies. First, as the MLA deforms the tension in the plantar fascia increases and excessive tension may lead to plantar fasciitis[21,23] and hallux rigidus/hallux limitus.[11] Second, change in navicular height was highly correlated ($r = 0.8$) with calcaneal inversion and eversion,[22] and eversion was highly coupled with internal tibial rotation,[27] thus making the foot susceptible to injuries associated with abnormal tibial rotation or pronation. This coupling of the foot and leg via the ankle-joint complex may partially explain why greater navicular drop was found in medial tibial stress syndrome patients[2] and those with anterior cruciate ligament ruptures compared to controls.[3] While these retrospective studies do not provide direct evidence of a link between MLA deformation and lower extremity pathologies, the biomechanical changes resulting from MLA deformation may provide insight into etiology and prevention strategies.

Collectively these studies provide quantitative data describing MLA kinematics during level-ground walking and sit-to-stand activities. For a broader understanding of MLA kinematics, exploration of different types of ambulation and the impact of load carriage parameters may be important. The aim of the current study was to describe the deformation of the MLA during various activities of daily living. Specifically, navicular displacement and arch elongation were observed during standing, walking, stair ascent and descent while carrying no load, a front load and a backpack load. It was hypothesized that navicular displacement and arch length would increase with an external load. It was also hypothesized that navicular displacement and arch length would increase with stair ambulation as compared to both static standing and walking.

METHODS

Participants

Young healthy adults (9 males, 8 females; age (mean ± SD): 26 ± 3 years; height: 1.72 ± 0.09 m; mass: 68.5 ± 9.7 kg; foot length: 250 ± 20 mm; foot breadth: 97 ± 8 mm; malleolus height: 71 ± 7 mm; malleolus width: 72 ± 5 mm) from the university population were recruited for this study. Potential participants were excluded if they had a history of musculoskeletal or neurological condition that would preclude safe walking and stair ambulation while carrying a moderate load, had current lower extremity discomfort, or wore foot orthotics. Experimental procedures were approved by the Iowa State University Institutional Review Board and all participants provided written informed consent prior to participation.

Apparatus

The experimental walkway for the walking trials was 5 m in length and the experimental staircase consisted of two steps (step height 18.5 cm, tread length 29.5 cm) and a top landing. During the front load trials, participants held a two-handled crate (20x31x20 cm, 13.6 kg total load). This mode of front load was chosen as it is more commonly encountered in the workforce than a front pack, and the selected mass is within the range (12.5 to 20 kg) previously used in other weighted-carry studies.[1,19] During the backpack load trials, participants wore a custom fitted, internal frame backpack (Kelty, Boulder, CO, USA) with a total mass of 13.6 kg.

Kinematic and kinetic recordings were collected simultaneously from an eight-camera, three-dimensional motion analysis system (Vicon, Oxford, UK), with a resolution of at least 1 ± 1 mm. Ten millimeter diameter retro-reflective markers were placed on the malleoli, navicular tuberosity, dorsifoot, first metatarsal head (2 cm from supporting surface), and medial calcaneus (2 cm from the supporting surface) of the right foot according to Nielsen et al.[26] while participants stood in equal-weight bearing on a custom-built apparatus for foot measurements. The medial calcaneus marker was placed 3 cm and 4 cm anterior from the most posterior aspect of the foot for females and males, respectively.[2] A posterior heel marker was placed in the center of the Achilles tendon, 2 cm from the supporting surface. The first metatarsal, navicular and calcaneus represent separate segments of the foot that experience significant sagittal plane rotation proposed by other researchers.[25,35]

Compared to radiographs, validity intraclass correlation coefficients range from 0.712 to 0.765 and 0.874 to 0.918 for truncated foot length and navicular height, respectively, measured using calipers with a resolution of 1 mm, similar to the anthropometry set used in the current study to mark the skin where the reflective markers would be placed.[34] While the variables and measurement tools are slightly different than the current study, similar validity may be expected. Measurement error due to skin artifact over the first metatarsal head, navicular and medial calcaneus range from 1.2 to 3.3 mm.[30]

Additional 1.9 cm diameter retro-reflective markers were placed on the malleoli, heel, dorsifoot and lateral aspect of the left foot, acromion processes, anterior and posterior superior iliac spines, greater trochanters, anterior thighs, medial and lateral femoral condyles, anterior shanks and sacrum.

Arch length was the three-dimensional distance from the medial calcaneus marker and first metatarsal head marker. Lengthening of the arch is expressed as a positive value from the reference unloaded standing trial. Navicular height was the three-dimensional, perpendicular distance from the navicular tuberosity marker to the line segment defined by the two endpoints of arch length (Figure 1). Change in navicular height was termed navicular displacement because it represents a three-dimensional movement of the navicular. It was also relative to the unloaded standing trial. A smaller distance from the navicular to arch length line is expressed as a positive value. Collectively, therefore, greater arch collapse is a positive value. Two in-ground and two portable force platforms (AMTI, Watertown, MA, USA) positioned on the first and second stair steps captured ground reaction forces. Kinematic data and ground reaction forces were sampled at 160 Hz and 960 Hz, respectively.

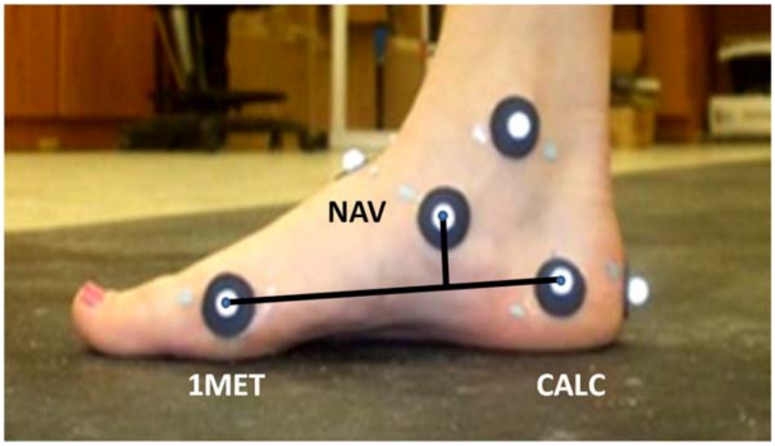

Figure 1. Arch length and navicular height (CALC: medial calcaneus, 1MET: first metatarsal head; NAV: navicular tuberosity).

Experimental Design

There were two independent variables in this study: MOTION and LOAD. The three levels of MOTION were: walking, stair ascent, and stair descent. The three levels of LOAD were: no load, front load (13.6 kg) and backpack load (13.6 kg) for a total of nine conditions (a 3×3 randomized complete block design with participants acting as blocks).

There were six dependent measures: 1) peak arch length elongation, 2) time to peak arch length elongation, 3) peak navicular displacement, 4) time to peak navicular displacement, 5) peak vertical ground reaction force, and 6) time to peak vertical ground reaction force. Times were expressed as a percent of the stance phase.

Procedure

In order to accurately measure the movement of the arch, the participants remained barefooted throughout the experiment. The gathered anthropometrics included height, whole body mass, right foot length, breadth, malleolus height and malleolus width. Markers were then placed according to the locations outlined in Apparatus above.

Testing began with three upright standing trials followed by a randomized presentation of each condition (five trials per condition). All participants were instructed to walk, ascend and descend stairs at a self-selected pace. During front load trials, participants held a crate (front load) at navel height, resting it against their body if desired. To reduce fatigue, participants were relieved of the front load between trials. After completing all experimental trials, two upright standing trials were performed to note if there were time-dependent changes in the MLA variables.

Data Processing and Analysis

All data were analyzed during the stance phase of walking and on the second step of both stair ascent and descent (right foot in all cases). A fourth order, symmetric Butterworth filter was applied to the video data with a cutoff frequency of 6 Hz, and the same filter was applied to the ground reaction force data with a 20 Hz cut-off frequency. The values for the five trials for an experimental condition were averaged for each participant and the means and standard deviations of all participants were calculated. Time at which peak values occurred was expressed as a percentage of the stance phase.

Statistical analysis was performed using SAS (Version 9.2, SAS, Cary, NC, USA). Prior to any statistical analysis, the assumptions of the ANOVA procedures (homogeneity of variances of the residuals, residuals normally distributed, and independence of observations) were evaluated using the techniques advocated by Montgomery.[24] Once these assumptions were verified, a MANOVA was performed to evaluate the effects of the independent variables on the set of dependent variables collectively, thereby controlling for the experiment-wise error rate. Univariate ANOVA procedures were then conducted and a Tukey-Kramer post-hoc test was performed to further explore the significant effects. The criteria p-value of $p < 0.05$ was used for all statistical tests.

RESULTS

The results of the MANOVA showed that both MOTION and LOAD had significant effects and were considered in subsequent univariate analyses (Table 1), while the MOTION×LOAD interaction was not significant ($p = 0.25$) and was not considered further. Subsequent univariate ANOVA further explored these main effects and showed that MOTION had a significant effect on both the magnitude of the peak navicular displacement (Figure 2) and the magnitude of the peak arch length elongation (Figure 3). Arch lengthening was greatest during walking and stair descent, and navicular displacement was greatest during stair descent. MOTION also had a significant effect on the timing of the peak navicular displacement, with peak displacement occurring at 60% of stance phase in stair ascent and 75 to 76% during walking and descent (Table 2). Exploration of the effect of LOAD showed that the peak vertical ground reaction force was significantly lower in the no load condition than either the front load or the back load condition. Finally, LOAD also had a significant effect on the timing of the peak vertical ground reaction force with the peak for stair descent occurring much earlier than either walking or stair ascent (Figure 4).

Table 1: MANOVA and Univariate ANOVA Results

Independent Variables	MANOVA	Peak Navicular Displacement	Time to Peak Navicular Displacement	Peak Arch Lengthening	Time to Peak Arch Lengthening	Peak Vertical Ground Reaction Force	Time to Peak Vertical Ground Reaction Force
MOTION	<0.0001	<0.0001	<0.0001	<0.0001	0.2417	<0.0001	<0.0001
LOAD	<0.0001	0.4569	0.5742	0.3258	0.2685	<0.0001	<0.0001

Table 2: Time to Peak Medial Longitudinal Arch Deformation for MOTION Type (Mean ± SD)

Independent Variable	Walking	Stair Ascent	Stair Descent
Time to peak navicular displacement (% stance)	76 ± 5	60 ± 20	75 ± 13
Time to peak arch length elongation (% stance)	46 ± 25	42 ± 20	50 ± 23

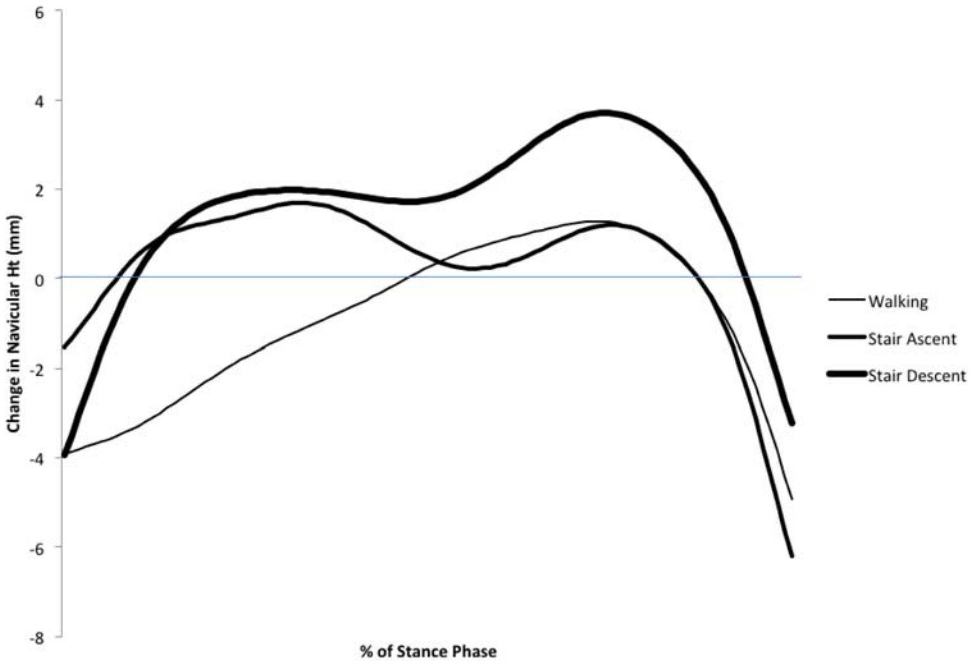

Figure 2. Representative navicular displacement during stance phase of walking, stair ascent, and stair descent for one trial for one participant. Zero displacement represents the same navicular height as during the static standing trial. A positive displacement represents a lower navicular height than standing and a negative

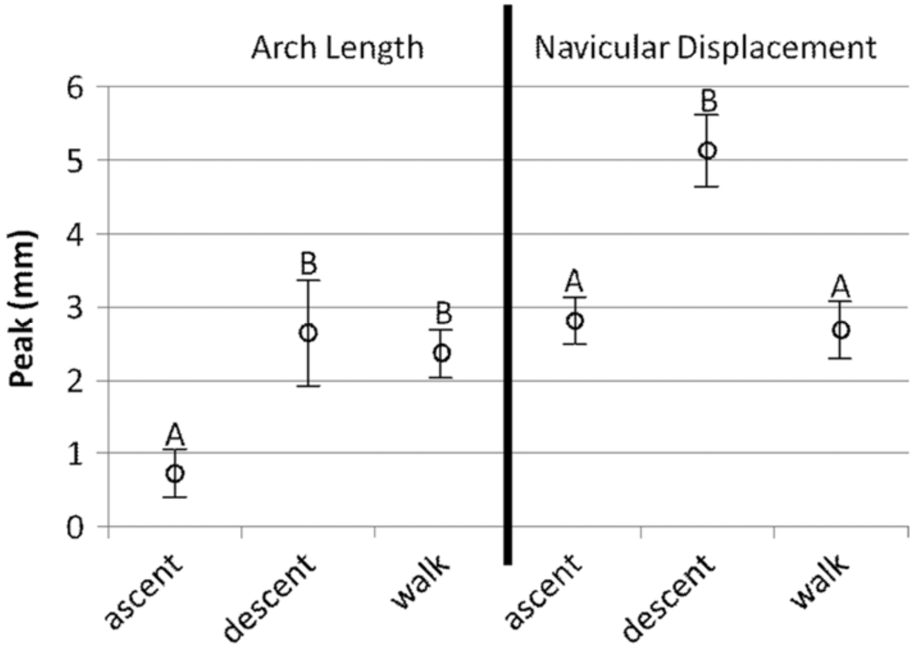

Figure 3. Mean (with 95% confidence interval) arch length elongation and navicular displacement as a function of MOTION type collapsed across load. Zero represents the same arch length and navicular height as during the static standing trial. Different letters indicate levels that are significantly different (e.g., navicular displacement during descent was greater than both ascent and walking, but ascent and walking are not statistically different from each other).

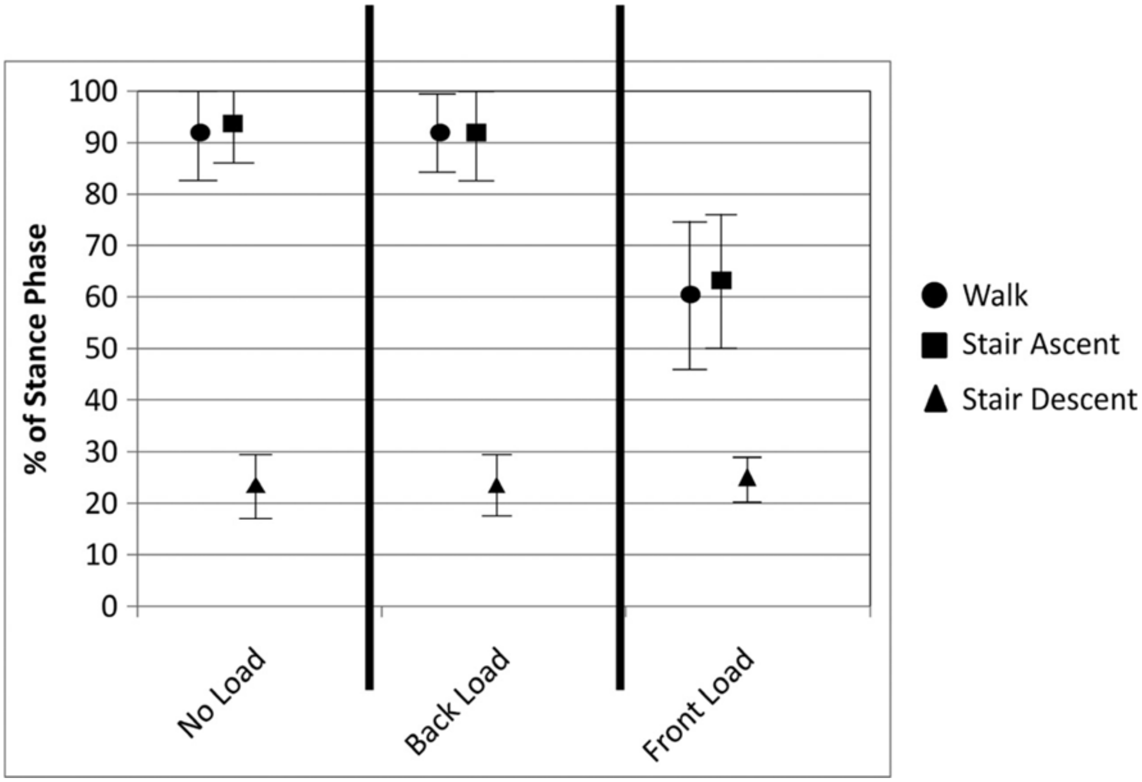

Figure 4. Timing of peak vertical ground reaction force as a function of MOTION and LOAD. The 95% confidence intervals are shown.

DISCUSSION

The purpose of this study was to expand our understanding of the MLA deformation response by quantifying it in a new set of activities of daily living: carrying loads and navigating stairs. Deformation of the MLA is a natural motion of the foot that absorbs energy through rotation of bones and deformation of tissues, such as the plantar fascia, plantar ligaments, and muscles/tendons.[18] It is important to quantify this motion in activities of daily living because excessive collapse may be related to injuries.[2,3,11,21,23] The MLA deformations experienced during the walking trials were consistent with those previously documented.[6,26] Navicular displacement was greatest in the stair descent task as compared to walking and ascent, and arch length elongation was significantly less in stair ascent as compared to level walking and descent. The changes in arch length observed in the current study ranged from 1.3 to 3.7%, which is similar to 2% reported by Cashmere et al.[6] during walking, but considerably less than the 9 to 12% reported by Gefen.[12] Various measurement techniques, reference lengths, definitions of arch length and arch height are likely to account for some of the variation in results among these studies. The magnitude of arch elongation observed was well below the threshold for acute strain damage previously estimated at 9%.[33] While greater strain leads to greater tension, which may be related to plantar fasciitis, Wearing et al.[32] did not observe any differences in sagittal plane MLA motion during walking between plantar fasciitis patients and controls. Therefore, it may be the repetitive loading and strain rate rather than the magnitude of the strain that increases injury risk.

In addition to the magnitude of these deformations, also of interest was the time at which these peaks occurred. The time to peak navicular displacement during walking was within the range of 50 to 85% of stance reported previously.[6,7,16] During walking and stair descent, maximum navicular displacement occurred at about 76% of stance, which was significantly later than stair ascent (60% of stance). This difference may be associated with achieving a more rigid, supinated foot posture earlier in stance in preparation for propulsion up the stairs and decreased sagittal ankle ROM during ascent compared to walking or descent. Maximum change in arch length generally occurred earlier in stance compared to peak navicular displacement, occurring at 42 to 50% of stance throughout all dynamic conditions. There was also an interesting response for the timing of the peak ground reaction force, illustrated in Figure 4. In the stair descent condition the peak vertical ground reaction force always occurred early in the stance phase, reflecting the significant weight-acceptance force at or near foot contact. In both walking and stair ascent conditions, the front load generated an earlier timing for the peak ground reaction force indicating that the anterior movement of the center of mass of the load-body system may result in a more rapid weight acceptance – a result consistent with Hsiang and Chang.[14]

The effects of the external load were not quite as clear and in some cases were inconsistent with our hypotheses. To begin, the effects of external load on peak ground reaction forces were consistent with previous research:[14] external load caused an increase in peak ground reaction forces for all levels of MOTION (as compared to the no load conditions). Interestingly, LOAD did not significantly affect any of the MLA deformation parameters, which did not support our original hypothesis. Since a previous study did find increased second metatarsal bone strain when carrying a 20 kg backpack,[1] perhaps our external load was not sufficient to cause additional deformation of the MLA. However, the external load (134 N) was greater than the increase in vertical ground reaction force observed during unloaded walking (~80 N), both compared to standing body weight. Under the dynamic conditions of the current experiment, it is difficult to separate the effects of motion and the effects of load because the dynamics of stair ascent/descent generate vertical inertial forces as compared to level walking. The variability in these inertial forces may have been of sufficient magnitude to obfuscate the underlying effects of the 13.6 kg external load. It is also possible that participants may have compensated at proximal joints to minimize arch deformation from the external load, such as increased hip ROM as has been observed during walking with a 12.5 kg vest.[19]

The timing of the lengthening and lowering of the arch did not always follow a consistent pattern relative to the timing of the peak ground reaction force. During walking and stair ascent, peak ground reaction force and navicular displacement typically occurred within 5% of each other. In contrast, during stair descent the peak ground reaction force occurred at approximately 23% of stance phase but maximum arch length elongation and navicular displacement occurred at 50% and 76% of stance, respectively. Therefore, peak navicular displacement may correspond more closely to peak plantarflexion moment, which another study found to occur right before toe-off during stair ambulation.[20] However, plantarflexion moment does not explain everything as Lin et al.[20] observed greater ankle moment in ascent versus descent, but we observed greater arch deformation in descent versus ascent. This must be interpreted with caution, though, as Lin et al.[20] did not simultaneously measure arch deformation.

Limitations relative to the generalizability of these results should be noted. First, participants in this study were relatively young, healthy college-age individuals and extrapolation to older persons or those with structural variances may be problematic. This is illustrated by navicular displacements during walking in the current study being less than that reported by Cornwall and McPoil.[7] A second limitation is the participants performed all trials barefoot, which may have caused changes in the normal lower extremity kinematics employed when shod. However, barefoot was necessary in order to track MLA motion using the motion analysis system. The use of reflective markers to measure MLA motion may have resulted in skin artifact of approximately the same magnitude as the changes observed,[30] but random error of measurement and relatively constant skin movement between conditions may affect the absolute values but not the differences between conditions. Additionally, we are assured MLA deformation occurs during dynamic activity based on studies using fluoroscopy and bone pins.[12,25] Future research may seek to use alternative motion tracking approaches that may allow for a verification of the results of the current study.

CONCLUSIONS

A 13.6 kg external load (backpack or front load) did not affect MLA deformation during walking or stair ambulation. Increases in arch length and navicular displacement were noted in walking and stair ambulation with stair descent generating the greatest deformations. Future research may seek to quantify other variables (center of pressure, ankle joint moments, muscle activation profiles, joint range of motion) during stair ambulation and walking to further explain MLA deformation patterns.

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