Available online at www.sciencedirect.com
ScienceDirect
Gait & Posture 28 (2008) 427–433
GAIT & POSTURE
www.elsevier.com/locate/gaitpost

Ankle fixation need not increase the energetic cost of human walking

Matthew T. Vanderpool [a,b], Steven H. Collins [b], Arthur D. Kuo [a,b,*]

[a] *Department of Biomedical Engineering, University of Michigan, Ann Arbor, MI, USA*
[b] *Department of Mechanical Engineering, University of Michigan, Ann Arbor, MI, USA*

Received 12 September 2007; received in revised form 2 December 2007; accepted 29 January 2008

Abstract

We tested whether the metabolic energy cost of walking with the ankles immobilized can be comparable to normal walking. Immobilization of any lower extremity joint usually causes greater energy expenditure. Fixation of the ankle might be expected to eliminate the work it normally performs, to detrimental effect. But fixation using lightweight boots with curved rocker bottoms can also bring some benefits, so that the overall energetic effect might be quite small. We measured oxygen consumption, kinematics, and ground reaction forces in six ($N = 6$) able-bodied human volunteers walking at 1.25 m/s in three conditions: normal walking in street shoes, walking with ankles immobilized by walking boots, and normally with ankles free but also weighted to match the mass of the walking boots. We estimated metabolic energy expenditure, joint work, and overall work performed on the body center of mass as a function of ankle fixation. Ankle fixation with walking boots caused the total rate of energy expenditure for walking to increase by 4.1% compared to normal ($P = 0.003$), but differed by an insignificant amount (0.4% less, $P = 0.78$) compared to walking with equivalent ankle weight. Compared to normal walking, ankle fixation can reduce ankle torque and work during the stance phase, most notably during late stance. This apparently makes up for the loss of ability to push-off as normal. With a suitably lightweight apparatus and curved rocker bottom surface, loss of ankle motion need not increase energy expenditure for walking.

Keywords: Gait; Biomechanics; Metabolic energy; Efficiency; Ankle fixation; Arthrodesis

1. Introduction

Fixation of the ankle is usually detrimental to walking economy. Compared to normal walking, ankle immobilization has been reported to result in greater metabolic energy expenditure, whether following arthrodesis [1], amputation [2], or fitting of ankle braces [3], walking casts [4,5], or orthoses [5]. For walking at the same speed, the increase is 9–15% [3,5]. Subjects with fixed ankles typically choose to walk more slowly than normal and expend about the same amount of energy, with a cost of transport (energy divided by weight and distance traveled) increasing by about 10% following arthrodesis [1,6], and 26% for normals wearing an ankle cast. A large increase in energy expenditure might be expected, given the ankle's prominent role in walking. The peak plantarflexion torque produced during stance is about double that of any other active joint torque [7]. The ankle also produces the largest burst of positive joint power, also with a peak about double that of the knee and hip. Given these contributions, it is perhaps curious that ankle fixation causes energy expenditure to increase by as little as 10%.

The intact ankle appears to operate quite economically, but this function may nonetheless come with costs. Humans normally produce active plantarflexion torque in the ankle during the second half of the stance phase [7]. This torque performs negative and then positive work about the ankle. Although some of this work may be performed elastically [8], there must be muscular effort associated with the plantarflexion torque and active work. This effort may be energetically costly despite resulting in relatively little network over a stride [9].

In terms of kinematics, the entire foot–ankle–shank complex appears to function very similarly to a rigid body

* Corresponding author at: 2350 Hayward Street, University of Michigan, Ann Arbor, MI 48109-2125, USA. Tel.: +1 734 647 2505; fax: +1 734 615 6647.
E-mail address: artkuo@umich.edu (A.D. Kuo).

0966-6362/$ – see front matter
doi:10.1016/j.gaitpost.2008.01.016

with a curved surface rolling on the ground. Observations of the forward progression of the center of pressure on the ground show that normal walking resembles rolling with a radius of curvature equal to about 30% of leg length [10,11]. Such a curved surface might be a suitable substitute for the intact ankle: it is standard clinical practice to restore walking function following ankle fusion with rigid-sole rocker bottom shoes [12]. Rocker bottoms have been demonstrated to redistribute plantar foot pressures [13,14] and improve gait kinematics [15] and functional performance [16]. Their effect on energy expenditure, however, has yet to be examined.

We previously studied how different rollover curvatures affect walking with ankles fixed (see Fig. 1a), examining how they affect the mechanical work performed on the body center of mass (COM) and the overall metabolic energy expended by able-bodied human participants [17]. We added

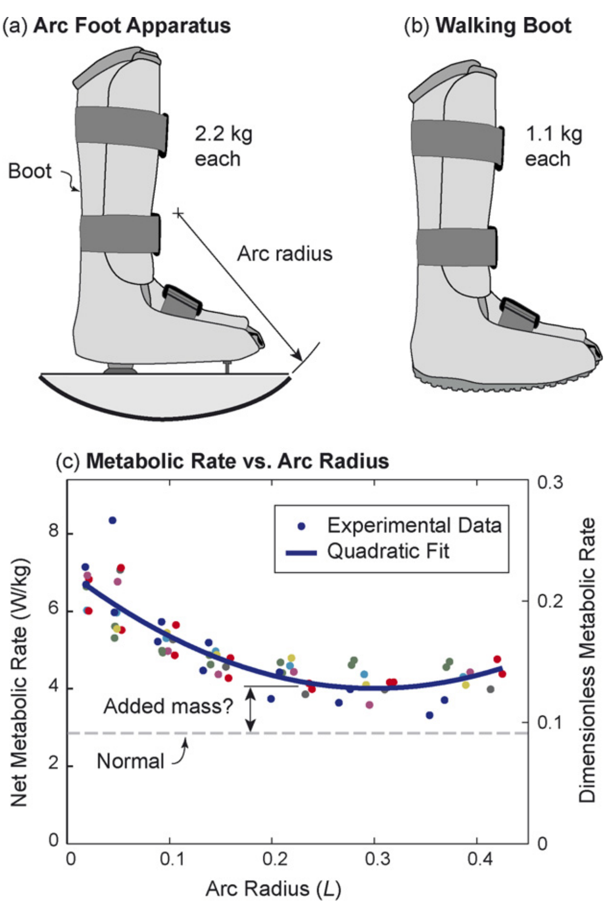

Fig. 1. Ankle fixation with different arc-shaped rocker bottoms. (a) Arc radius was previously varied by attaching rigid rocker bottoms to a modified pair of walking boots [17]. (b) Unmodified Aircast (Summit, NJ) Pneumatic Walker boots also have rigid rocker bottoms. (c) Net metabolic energy expenditure (cost of walking minus that for quiet standing) varied with arc radius, with a minimum at a radius of about 0.3 (as fraction of leg length, L). The difference above normal was potentially explained by the added mass of the arc foot apparatus, suggesting that ankle fixation need not lead to greater energy expenditure for walking. The right-hand axis is labeled with dimensionless units.

different arc-shaped rocker bottom surfaces to a pair of walking boots that fixed the ankle joints (Fig. 1b). Curvatures close to 30% of leg length allowed participants to walk at the same speed as normal but with decreased mechanical work performed on the COM by the individual legs. Less work did not, however, result in lower metabolic energy expenditure. Expenditure actually increased compared to normal (Fig. 1c), more consistent with other reports concerning ankle fixation [5]. But the increase we observed was within the range expected from walking with added mass at the ankles, due to the weight of the apparatus we used to fixate the ankles and apply the rocker bottom curvature [17]. Given a proper rocker bottom shape, it appears possible for ankle fixation to have no detrimental effect on energy expenditure other than added mass.

The purpose of the present study was to compare walking with fixed ankles against normal walking while controlling for added mass. We expected that walking with ankles fixed by a suitably lightweight apparatus with comfortable rolling surface would have some advantages over normal walking, such as decreased need to produce active ankle torque and to perform work on the COM. On the other hand, we also expected that normal ankle function would have advantages that justify its use in normal walking. It is unclear, however, which set of advantages must prevail. Our previous observations [17] led us to expect very little difference in energy expenditure between the two cases.

2. Materials and methods

We compared the energetics and mechanics of able-bodied human participants walking with normal ankle function versus walking with ankles bilaterally fixed by walking boots. We measured oxygen consumption to quantify metabolic energy expenditure, and joint powers and work performed on the COM to quantify gait mechanics. The comparisons were all made for a single walking speed but in two sub-studies, one with energetics data recorded during treadmill walking and the other with mechanics measurements conducted during overground walking. The effect of the mass of walking boots was also examined, through the addition of ankle weights to otherwise normal walking.

A total of 11 adult male participants (aged 22–40 years) participated in the entire study. All participants provided informed consent according to University Institutional Review Board Safety Procedures, and all were considered to be nonpathologic ambulators in good health. Six ($N = 6$, body mass 80.2 ± 7.0 kg, leg length 0.915 ± 0.058 m, mean ± S.D.) of the participants participated in the energetics trials, and eight ($N = 8$, body mass 75.9 ± 7.7 kg, leg length 0.934 ± 0.073 m) participated in the mechanics trials (with three subjects participating in both). All walking trials were conducted at a speed of 1.25 m/s.

Energetic cost was measured for three walking conditions, comparing normal walking against walking with ankles fixed and normal walking with ankle weights, hereafter referred to as Normal, Ankles Fixed, and Ankles Weighted conditions, respectively. Normal walking trials were conducted while participants wore the street shoes they were using that day (mass 411 ± 34 g each). Ankles were immobilized using Aircast Pneumatic Walker (Aircast Inc., Summit, NJ, USA) boots size large, hereafter referred to simply as ''walking boots'' (mass 1120 ± 3 g each). These are commonly prescribed to fixate ankles following ankle and foot injuries, and were selected because they are lightweight and provide a curved rocker bottom surface that appears comfortable to walk on. The boots contain air bladders that help to immobilize the foot; these were inflated according to manufacturer guidelines. During Weighted conditions, participants wore adjustable ankle weights (Style #300 ankle cuffs, All Pro Exercise Products Inc., Longboat Key, FL, USA). Appropriate amounts of mass were added, in increments as small as 10 g, so that the total mass including street shoes was close to that of the walking boots (mean total mass for condition 1123 ± 17 g). This accounted for the mass but not the higher rotational inertia of the boots.

For energetics trials, we measured the rate of oxygen consumption ($\dot{V}_{O_2}$ in ml O_2/s) and carbon dioxide production ($\dot{V}_{CO_2}$ in ml CO_2/s) using an open-circuit respirometry system (Physio-Dyne Instrument, Quogue, NY). Each trial lasted at least 7 min, including at least 3 min to allow participants to reach steady-state, followed by 3 min of data recording for average $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$ during steady-state. Metabolic rates $\dot{E}$ (in W) were estimated with the standard formula [18]:

$$\dot{E} = 16.48\,\dot{V}_{O_2} + 4.48\,\dot{V}_{CO}.$$

We also measured each subject's metabolic rate for quiet standing in a separate trial of the same duration, and subtracted it from the rate for walking to yield a net metabolic rate. All conditions, including quiet standing, were conducted in random order. Respiratory exchange ratios were less than unity for all participants and conditions, indicating that energy was supplied primarily by oxidative metabolism in all test conditions. No metabolic data were collected during overground walking.

For mechanics trials, we measured kinematics and ground reaction forces as participants walked overground-embedded force plates in the Ankles Fixed and Normal conditions. Kinematic data were recorded with an 8-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) at 120 Hz. Force data were recorded at 1200 Hz with two force plates (AMTI, Watertown, MA) placed in tandem. Speed was measured with two photogates, positioned 2.5 m apart; trials were discarded if actual walking speed was not within 5% of the desired speed of 1.25 m/s. We recorded five successful trials per condition for each subject. For inverse dynamics analysis, a set of 29 motion capture markers were placed bilaterally on the lower extremity. Marker locations comprised the fifth metatarsal of the foot, the heel, the medial and lateral malleolli, the medial and lateral epicondyles of the knee, the greater trochanter, the anterior superior iliac spine, the sacrum, and a three-marker cluster on each thigh and shank. Markers were placed in the same relative locations on the walking boots as they were on normal street shoes, with no different assumptions regarding ankle motion.

3. Analysis

We computed joint powers and the work performed on the COM for all conditions. Commercially available gait analysis software (C-motion Visual 3D: Rockville, MD) was used to calculate bilateral ankle, knee, and hip joint torques, powers, and work for each overground walking trial, taking into account the mass of the walking boots as appropriate. The velocities and torques were low-pass filtered at 12 Hz as part of this analysis.

We used ground reaction forces to compute the rate of work performed on the COM by each leg, defined as the vector dot product of each leg's ground reaction force against the COM velocity [19,20]. COM work is helpful for quantifying how much positive work is performed during push-off and how much negative work is performed during the collision of the leading (swing) leg with ground [20]. Joint work was computed for the sagittal plane, whereas COM work was computed in all three spatial dimensions. Each trial was normalized in duration to percent gait cycle and averaged for each subject. All torque, power, and work quantities were analyzed in dimensionless form, to help account for variations in subject size. Torque, power, and work quantities were normalized by each subject's body weight and leg length (MgL, where M is body mass, g is gravitational acceleration, and L is leg length), with the additional factor of $g^{0.5}L^{-0.5}$ (the leg's pendulum frequency) for power. Averages computed across participants used dimensionless variables. We also report most variables in more familiar dimensional units such as W kg^{-1} for metabolic rate, converted by multiplying each dimensionless variable against the appropriate average normalization factor.

We statistically compared several outcome variables that summarize the energetics and mechanics. We compared net metabolic rate, total positive and negative joint work per stride, and COM work during four gait phases. Joint work per stride was defined as the time-integral of the joint powers over the intervals during which they were positive, with a corresponding measure for negative joint work. COM work was defined as the integrated work during phases termed push-off, rebound, pre-load, and collision, delineated by the major zero-crossings of COM work rate [21]. For energetics data, statistical comparisons were made with repeated measures ANOVA for each variable, with a significance level α of 0.05. Where differences were significant, post hoc comparisons were performed using paired, two-tailed t-tests, controlling for family wise errors with a multiple-comparison α of 0.05. For mechanics data where there were only two conditions, we conducted paired, two-tailed t-tests with an α of 0.05.

4. Results

In normal walking trials, participants consumed an average of 16.0 ml O_2/s ($\pm$1.8 standard deviation, S.D.) and produced 12.7 ml CO_2/s ($\pm$1.6 S.D.). The corresponding quantities for quiet standing were 4.6 ml O_2/s ($\pm$1.0 S.D.) and 3.7 ml CO_2/s ($\pm$0.7 S.D.). The total metabolic rate for normal walking was therefore 4.0 ± 0.3 W kg^{-1} or 0.130 (dimensionless, $\pm$0.010 S.D.), with a net metabolic rate of 2.8 $\pm$0.2 W kg^{-1} or 0.092 (dimensionless, $\pm$0.007 S.D.).

Walking in the Ankles Fixed and Weighted conditions (see Fig. 2) resulted in small but statistically significant

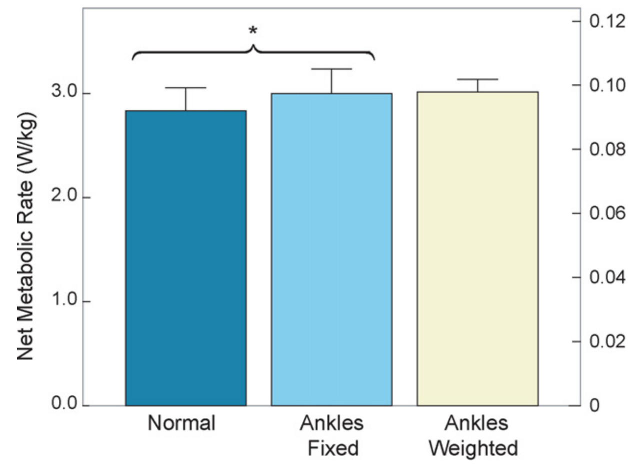

Fig. 2. Net metabolic rate for walking under Normal, Ankles Fixed, and Ankles Weighted conditions. The latter two conditions required significantly greater net energy expenditure (5.8% and 6.4%, respectively). There was no significant difference between Fixed and Weighted conditions. (* $P < 0.05$, post hoc paired t-test, $N = 6$.) The right-hand axis is labeled with dimensionless units.

increases in energy expenditure ($P = 0.014$, repeated measures ANOVA). The Fixed condition resulted in a 4.1% increase in total metabolic rate ($P = 0.0026$, post hoc paired, two-tailed t-test), and the Weighted condition in a 4.5% increase ($P = 0.0097$), compared to the Normal condition. There was not a statistically significant difference between Fixed and Weighted conditions ($P = 0.78$). The respective increases were 5.8% and 6.4% in terms of net metabolic rate.

Immobilization of the ankles also resulted in changes in joint kinematics, torques, and powers (see Fig. 3). Qualitatively, the changes were largest for the ankles and the smallest for the hips. In terms of kinematics, ankle motion with the Ankles Fixed was greatly restricted compared to Normal, but the knee and hip joint angles were within the normal range. Joint torques were quite similar, with the Ankles Fixed torques lying almost entirely within the Normal range, even though the ankles did not move. This lack of motion did, however, result in large differences in ankle powers (the product of joint torque and angular velocity), with close to zero power in the Fixed condition. The knee and hip powers remained almost entirely within the Normal range, except for a reduced magnitude of the negative knee power that is normally observed near the end of stance (50–60% of stride), during push-off. We also observed a brief increase in positive hip power at the end of double support (about 12% of stride).

Joint work summary variables showed statistically significant differences only for the ankle (see Fig. 4).

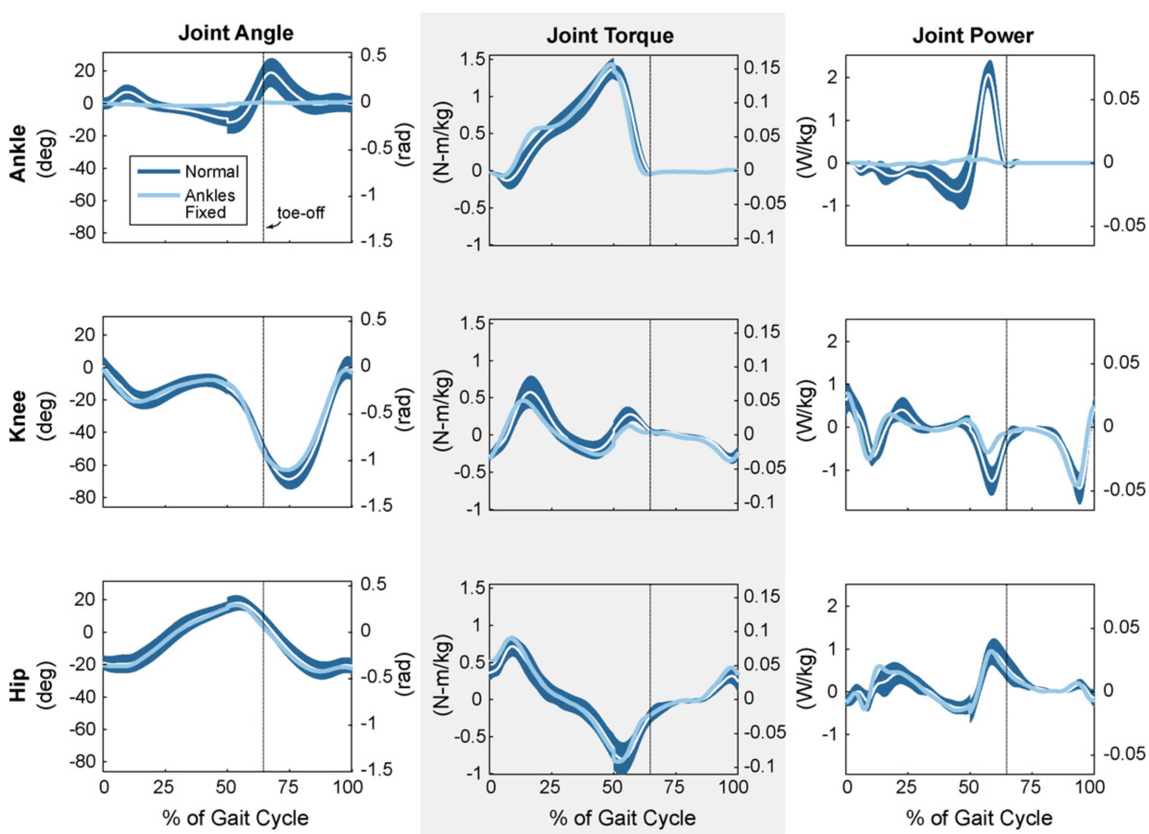

Fig. 3. Joint angles, torques, and powers over a stride, for ankle, knee, and hip joints. The range of Normal values, within 1 standard deviation of mean (shaded regions) across subjects is compared with the Ankles Fixed condition ($N = 8$). The greatest differences between conditions may be observed in ankle angles and powers, even though the torques are quite similar. Data shown are derived from one bilateral step, made to appear as one unilateral stride assuming lateral symmetry. The right-hand axes are labeled with dimensionless units. Angles and torques are defined as positive in the extension direction.

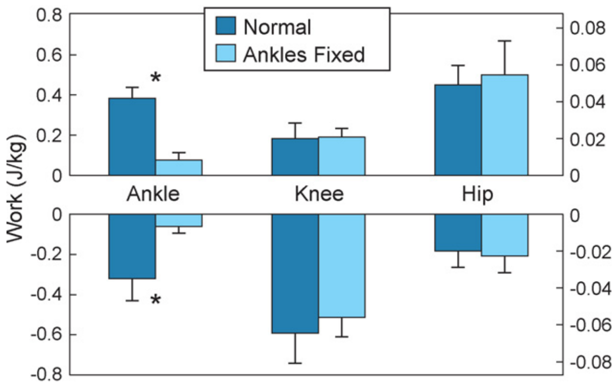

Fig. 4. Cumulative positive and negative work performed at the joints over a stride, in Normal and Ankles Fixed conditions. Positive work is defined as the joint power (see Fig. 3, right column) integrated cumulatively over the intervals where it is positive, complemented by a corresponding negative work integral. Only the work performed at the ankle was significantly affected by immobilization, decreasing by 80.5% and 81.6%, for positive and negative work, respectively. (* $P < 0.05$, paired t-tests, $N = 8$.) The right-hand axis is labeled with dimensionless units.

Positive ankle work (over a stride) in the Fixed condition was only 19.5% of Normal ($P = 4.4\text{E}{-5}$), and negative ankle work was only 18.4% of Normal ($P = 6.4\text{E}{-4}$). There were no statistically significant differences for the knee (positive work $P = 0.94$, negative work $P = 0.058$) or hip (positive work $P = 0.99$, negative work $P = 0.21$). In other words, decreased ankle in the Ankles Fixed condition was not compensated by greater work at other joints.

Immobilization of the ankles also resulted in changes in the work performed on the COM. Instantaneous COM work rate (see Fig. 5) varied over a stride with a similar pattern to Normal, but with generally lower amplitudes. The cumulative negative work performed on the COM over a stride was 0.036 ± 0.005 (dimensionless) in the Normal condition, and 0.0269 ± 0.004 with Ankles Fixed, a reduction of 25% ($P = 6.6\text{E}{-4}$). The positive work of push-off alone was reduced by about 35% ($P = 0.0013$). The negative work during the pre-load phase was reduced in magnitude by about 41% ($P = 0.0036$). The other phases did not exhibit statistically significant differences.

5. Discussion

The aim of this study was to test whether immobilization of the ankles must result in greater energy expenditure during walking. We proposed that, with a suitable rocker bottom shape, there need not be an increase in energy expenditure other than for the added weight of the immobilization device. Our results here showed only a 4.1% increase in total metabolic rate with ankles immobilized compared to normal walking, and no difference compared to walking normally but with an equivalent mass added to the ankles. This might appear to run counter to the previous literature, but perhaps only because other studies have not taken advantage of curved rocker bottoms with lightweight boots.

Our findings are consistent with those for walking with ankles similarly immobilized but with a variety of heavier, curved rocker bottom shapes [17]. Curvatures with radius about 30% of leg length resulted in decreased work performed on the COM compared to normal, mainly due to low heel strike collisions afforded by the curvature. The collision work may be predicted with remarkably simple models, and the COM work analysis is designed to measure and test these predictions [22], with joint powers indicating that much of the negative work is performed at the knee (see Fig. 3). Although collision work can predict one contribution to energetic cost, other costs also apply, such as for forced motion of the legs back and forth [23,24]. The latter is sensitive to added mass at the legs, possibly accounting for the greater net energy expenditure we observed with the heavy rocker bottom shapes. Here we used lightweight boots with a different curvature that nonetheless resulted in a similar amount of COM work (0.0269 dimensionless negative work per stride vs. 0.0272 previously [17]). These results indicate that added mass can account for a substantial fraction of overall energy expenditure.

The shape of the rocker bottom may also explain why other studies have reported higher energy expenditure with ankle immobilization [1,3–5]. Fixation, whether by a cast, splint, or arthrodesis, is often not accompanied by a favorable rolling shape and reduced collision losses. A less favorable shape may require more positive work at push-off or elsewhere in the stride to compensate for the loss. This work might not be performed with the same efficiency as normal ankle push-off [25,26]. Another potential explanation applies to studies that examined participants who had undergone actual arthrodesis or amputation [1,2]. These participants may have had comorbidities or other complications that adversely affected energy expenditure. Lightweight rocker bottom shapes and the testing of only able-bodied subjects could both contribute to lower energy expenditure in the present study.

Ankle fixation with walking boots appears to have some advantages compared to normal walking. Fixation allows the plantarflexion moment occurring after mid-stance to be supported passively rather than by active muscle. The reduced motion also means that little positive and negative work is performed at the ankle (see Fig. 3), during phases we label as Push-off and pre-load, respectively (see Fig. 5). Although much of the pre-load negative work may be restored elastically during push-off [8,9], we would nevertheless expect reduced ankle torque to result in decreased energy expenditure. Reduced push-off might be disadvantageous if negative Collision losses were higher, but the boots' curvature keeps these losses low. To the extent that normal ankle function effectively resembles the rolling of a rigid surface, it may be advantageous to substitute an actual rigid surface.

Of course, there may also be disadvantages to immobilization. It limits – but does not eliminate – the

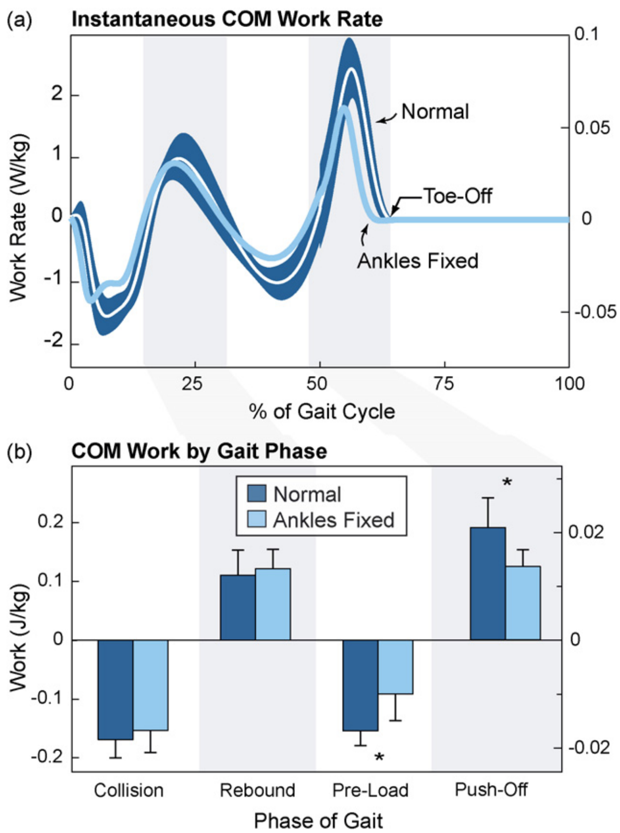

Fig. 5. Work performed on the COM, in terms of (a) instantaneous work rate as a percentage of stride cycle and the (b) cumulative work performed during each of four gait phases. Instantaneous work rate in the Ankles Fixed condition was generally of lower amplitude than Normal. The work may be separated into collision, rebound, pre-load, and push-off phases, demarcated by major zero-crossings so that the phases summarize alternating regions of mostly negative or positive work. The Ankles Fixed condition resulted in decreased magnitude of negative and positive work during the pre-load and push-off phases, respectively. (* $P < 0.05$, paired t-tests, $N = 8$.) The right-hand axes are labeled with dimensionless units.

body's ability to push off, and the substitution of muscles other than the ankle plantarflexors to achieve push-off may cause the work to be performed with lower efficiency than normal. Some rocker bottoms, particularly those with small radius of curvature, induce large collision losses [17], which require more positive work in compensation. Limited push-off means that some additional work may be required of other joints, perhaps at other times in the stride, and potentially with greater energy expenditure [17]. Push-off limitations did not appear problematic in the present study, perhaps owing to favorable rocker bottom shapes that required little push-off. Immobilization also limits ankle eversion/inversion [27], which may reduce the ability to control balance during walking, leading to compensations that may be energetically costly [28]. Intact ankle motion is also particularly helpful on uneven ground, for descending stairs, or performing other gaits. We also found the curved bottom of the walking boots to be subjectively less stable for upright standing than normal shoes. It is therefore mainly when walking on level ground that ankle immobilization appears to have little adverse effect.

There are a number of limitations to this study. We have not systematically studied rocker bottom shapes other than circular arcs and that of the walking boots. Both rocker bottom shape and length may potentially affect energy expenditure [29], and some combinations might actually improve on those that we have examined thus far. We also examined only one size of walking boots, with male participants of sufficient height and shoe size to fit those boots well. It would be desirable to perform additional measurements on a range of subject sizes and both genders to test the generality of the findings reported here. We also did not measure the actual ankle muscle activity or torque that may have been produced during Ankles Fixed walking. It is likely that participants produced some residual torque within the walking boots. Several participants reported subjectively that their muscle activity decreased as they became acclimated to the boots. It is possible that extended practice could result in decreased ankle muscle activity. It might also improve balance while wearing the boots. Optimization of rocker bottom geometry and greater practice time could potentially reduce energy expenditure for walking with ankle fixation. We also did not have the exact same subject group perform the mechanics and energetics trials, which were also conducted overground and on a treadmill, respectively. It would be preferable to use an identical subject group, although all comparisons were made only within each sub-study rather than between groups. There may also be small differences between overground and treadmill walking, which might have slight effect on the absolute amount of energy or work performed in a condition, but would not be expected to have substantial effect on the observed trends.

These limitations do not, however, detract from our finding that ankle immobilization need not lead to greater energy expenditure. The shape of the rocker bottom and the weight of the fixation apparatus both appear to influence energy expenditure. Ankle fixation may have disadvantages such as reduced push-off and poorer balance, but these are offset by advantages such as reduced need for active ankle plantarflexion torque or work, and reduced collision losses.

6. Conclusions

The energetic cost of walking with an immobilized ankle is statistically no greater than that for walking with a weight-matched but free ankle. Ankle fixation reduces the active plantarflexion work performed during late stance. Although less positive work can be performed in ankle push-off, a curved rocker bottom surface appears to also reduce negative work, so that less positive work need be performed on the body center of mass over a stride. This leads to only a 4.1% increase in total energy expenditure compared to normal walking, that can potentially be explained by the added mass of the fixation apparatus alone. The consideration of weight and curvature of rocker bottoms may be relevant to energy expenditure, with potential applications to ankle arthrodesis or amputation, or the fitting of orthoses.

Conflict of Interest Statement

S.H.C. is President of Intelligent Prosthetic Systems LLC, which develops prosthetic foot technology related to this study.

Acknowledgement

This work was supported in part by the National Institutes of Health (DC6466).

References

[1] Waters RL, Barnes G, Husserl T, Silver L, Liss R. Comparable energy expenditure after arthrodesis of the hip and ankle. J Bone Joint Surg Am 1988;70(7):1032–7.
[2] Waters RL, Perry J, Antonelli D, Hislop H. Energy cost of walking of amputees: the influence of level of amputation. J Bone Joint Surg Am 1976;58(1):42–6.
[3] Ralston H. Effects of immobilization of various body segments on the energy cost of human locomotion. Ergon Suppl 1965;53:39–46.
[4] Waters RL, Campbell J, Thomas L, Hugos L, Davis P. Energy costs of walking in lower-extremity plaster casts. J Bone Joint Surg Am 1982;64(6):896–9.
[5] Fowler PT, Botte MJ, Mathewson JW, Speth SR, Byrne TP, Sutherland DH. Energy cost of ambulation with different methods of foot and ankle immobilization. J Orthop Res 1993;11(3):416–21.
[6] Waters RL, Mulroy S. The energy expenditure of normal and pathologic gait. Gait Posture 1999;9(3):207–31.
[7] Winter DA. The biomechanics and motor control of human gait: normal, elderly and pathological, 2nd ed., Waterloo, Ontario: Waterloo Biomechanics; 1991.
[8] Lichtwark GA, Wilson AM. Interactions between the human gastrocnemius muscle and the Achilles tendon during incline, level and decline locomotion. J Exp Biol 2006;209(Pt 21):4379–88.
[9] Hof AL, Van Zandwijk JP, Bobbert MF. Mechanics of human triceps surae muscle in walking, running and jumping. Acta Physiol Scand 2002;174(1):17–30.
[10] Hansen A, Gard SA, Childress DS. The determination of foot/ankle roll-over shape: clinical and re-search applications. In: Harris G, Smith P, editors. Pediatric gait: a new millennium in clinical care and motion analysis technology. Piscataway, NJ: Institute of Electrical and Electronics Engineers Inc.; 2000.
[11] McGeer T. Passive dynamic walking. Int J Robotics Res 1990;9(2): 62–82.
[12] Wagner Jr FW. Ankle fusion for degenerative arthritis secondary to the collagen diseases. Foot Ankle 1982;3(1):24–31.
[13] Schaff PS, Cavanagh PR. Shoes for the insensitive foot: the effect of a "rocker bottom" shoe modification on plantar pressure distribution. Foot Ankle 1990;11(3):129–40.
[14] Shereff MJ, Bregman AM, Kummer FJ. The effect of immobilization devices on the load distribution under the foot. Clin Orthop Relat Res 1985;(192):260–7.
[15] Wu WL, Rosenbaum D, Su FC. The effects of rocker sole and SACH heel on kinematics in gait. Med Eng Phys 2004;26(8):639–46.
[16] Mueller MJ, Strube MJ. Therapeutic footwear: enhanced function in people with diabetes and transmetatarsal amputation. Arch Phys Med Rehabil 1997;78(9):952–6.
[17] Adamczyk PG, Collins SH, Kuo AD. The advantages of a rolling foot in human walking. J Exp Biol 2006;209:3953–63.
[18] Brockway JM. Derivation of formulae used to calculate energy expenditure in man. Hum Nutr Clin Nutr 1987;41(6):463–71.
[19] Donelan JM, Kram R, Kuo AD. Mechanical and metabolic determinants of the preferred step width in human walking. Proc R Soc Lond B Biol Sci 2001;268(1480):1985–92.
[20] Donelan JM, Kram R, Kuo AD. Simultaneous positive and negative external mechanical work in human walking. J Biomech 2002;35(1): 117–24.
[21] Donelan JM, Kram R, Kuo AD. Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. J Exp Biol 2002;205(Pt 23):3717–27.
[22] Kuo AD. The six determinants of gait and the inverted pendulum analogy: a dynamic walking perspective. Hum Mov Sci 2007;26: 617–56.
[23] Doke J, Donelan JM, Kuo AD. Mechanics and energetics of swinging the human leg. J Exp Biol 2005;208(Pt 3):439–45.
[24] Doke J, Kuo AD. Energetic cost of producing muscle force, rather than work, to swing the human leg. J Exp Biol 2007;210(Pt 13):2390–8.
[25] Kuo AD. A simple model of bipedal walking predicts the preferred speed-step length relationship. J Biomech Eng 2001;123(3):264–9.
[26] Kuo AD. Energetics of actively powered locomotion using the simplest walking model. J Biomech Eng 2002;124(1):113–20.
[27] Hof AL, van Bockel RM, Schoppen T, Postema K. Control of lateral balance in walking. Experimental findings in normal subjects and above-knee amputees. Gait Posture 2007;25(2):250–8.
[28] Donelan JM, Shipman DW, Kram R, Kuo AD. Mechanical and metabolic requirements for active lateral stabilization in human walking. J Biomech 2004;37(6):827–35.
[29] Ruina A, Bertram JE, Srinivasan M. A collisional model of the energetic cost of support work qualitatively explains leg sequencing in walking and galloping, pseudo-elastic leg behavior in running and the walk-to-run transition. J Theor Biol 2005;237(2):170–92.