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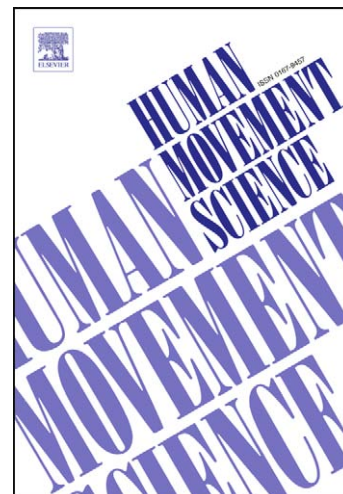
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Metabolic cost and mechanical work for the step-to-step transition in walking after successful total ankle arthroplasty

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The study was approved by the institutional review board of the Slotervaart Hospital and all participants gave informed consent prior to data collection.

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Abstract

The aim of this study was to investigate whether impaired ankle function after total ankle arthroplasty (TAA) affects the mechanical work during step-to-step transition and the metabolic cost of walking. Respiratory and force plate data were recorded in 11 patients and 11 healthy controls while they walked barefoot at a fixed walking speed (FWS, 1.25 m/s) and at their self-selected speed (SWS). At FWS metabolic cost of transport was 28% higher for the TAA group, but at SWS there was no significant increase. During the step-to-step transition, positive mechanical work generated by the trailing TAA leg was lower and negative mechanical work in the leading intact leg was larger. Despite the increase in mechanical work dissipation during double support, no significant differences in total mechanical work were found over a complete stride. This might be a result of methodological limitations of calculating mechanical work. Nevertheless, mechanical work dissipated during the step-to-step transition at FWS correlated significantly with metabolic cost of transport: $r = .540$. It was concluded that patients after successful TAA still experienced an impaired lower leg function, which contributed to an increase in mechanical energy dissipation during the step-to-step transition, and to an increase in the metabolic demand of walking.

Keywords: Energy Consumption; Mechanical Work; Barefoot Walking, Total Ankle Arthroplasty.

1. Introduction

The foot and ankle complex is a versatile complex of joints and muscles that forms a link between the floor and the body in bipedal movement tasks, such as walking. Due to its role as the interface between body and surroundings its behavior has large consequences for the behavior of the entire body on top. In case of pathology of the foot-ankle complex, this will have a pronounced influence on walking ability.

People with end-stage arthritis of the ankle joint are often treated with a surgical fixation of this joint (i.e., ankle arthrodesis) as a final clinical intervention to relieve the pain and restore functionality. This intervention has been shown to result in a reduced walking speed and step length (Waters & Mulroy, 1999, Wu et al., 2000), movement compensation in the adjacent tarsal joints (Wu et al., 2000) and increased energy demand during walking (Waters, Campbell, Thomas, Hugos, & Davis, 1988). As an alternative to surgical fixation, in this group of patients ankle motion can be partially preserved with the use of an endoprosthesis, i.e., total ankle arthroplasty (TAA). In TAA the arthritic surfaces of the distal tibia and talar dome are replaced by either semiconstrained two-component fixed-bearing designs or unconstrained three-component mobile-bearing designs. Good medium-term clinical results of TAA with use of such designs have been described in recent years (Doets, Brand, & Nelissen, 2006; Knecht et al., 2004; Kofoed & Lundberg-Jensen, 1999; Wood, Prem, & Sutton, 2008). Although endoprostheses have been shown to preserve ankle motion during walking (Doets, van Middelkoop, Houdijk, Nelissen, & Veeger, 2007), walking speed and mechanical power output of the ankle joint remain reduced (Dyrby, Chou, Andriacchi, & Mann, 2004; Houdijk, Doets, van

Middelkoop, & Veeger, 2008). The effect of TAA on the metabolic cost of walking is yet unknown.

The metabolic cost of walking is regarded to be an important characteristic of gait, which determines the ability of people to engage in prolonged walking activities (Waters & Mulroy, 1999). Restriction of ankle function has been shown to affect metabolic cost of walking both in practice and theory. From experimental research it is known that restriction of ankle motion by an external immobilization (Fowler et al., 1993; Waters, Campbell, Thomas, Hugos, & Davis, 1982) or through ankle arthrodesis (Waters et al., 1988) raises energy consumption during walking. However, a satisfying biomechanical explanation for this phenomenon remains to be given.

A theoretical model on the role of ankle plantar flexion on metabolic cost of walking was recently presented by Kuo, Donelan, and Ruina (2005). This so-called double inverted pendulum model describes and predicts the mechanical work necessary for the transition from one step to the next in walking. In its simplest form the human body can be modeled as two rigid legs with a point mass on top. During the step-to-step transition of this model, the center of mass (CoM) velocity has to be redirected from a circular trajectory around the trailing leg to a new circular trajectory around the leading leg. Redirection of the CoM can occur through an impulsive force generated by the leading leg during heel contact. However, this impact force dissipates mechanical energy. To walk at a steady velocity this negative work has to be restored. The double inverted pendulum model predicts that restoring mechanical work could best be done by the trailing leg at the instant of heelstrike through a powerful plantarflexion (Kuo, 1999, 2002; Kuo et al., 2005). Alternatively, positive work could be generated substantially prior to heelstrike by

torques generated around the hip. The model predicts however that this “hip strategy” would result in an increase in negative work during collision, and consequently in an increase in the mechanical work necessary for the step-to-step transition, which would induce a higher metabolic energy cost for walking. This mechanism could explain the higher metabolic energy consumption found in people walking with a restricted ankle function, who are then forced to use a hip strategy.

In this study we set out to investigate the metabolic cost and the mechanical work performed on the CoM during walking in people after TAA, in order to find whether their impaired ankle function results in an increase in the mechanical work for the step-to-step transition during walking and whether this coincides with an increased metabolic cost. These results will contribute to the validation of the double pendulum model of walking and to our understanding of the crucial role of the ankle in locomotion, as well as to our understanding of the clinical implications of TAA on the metabolic cost of walking.

2. Materials and methods

2.1. Participants

The study was approved by the institutional review board and all participants gave informed consent prior to data collection. Two groups were included: a control group and a patient group with a successful mobile-bearing TAA. The control group consisted of 11 healthy participants without any impairment of the lower extremities. Inclusion criteria for the TAA group were: a primary diagnosis of osteoarthritis or rheumatoid arthritis of the ankle joint; a good clinical outcome as defined by ankle scores of more than 80 points

on both the Low Contact Stress (LCS) ankle score (Buechel, Pappas, & Iorio, 1988) and the American Orthopaedic Foot and Ankle Society (AOFAS) ankle-hindfoot score (Kitaoka et al., 1994). Furthermore, alignment of the ankle-hindfoot complex should be neutral and range-of-motion, as measured by manual goniometry, should be a minimum of 10 degrees of dorsiflexion and 20 degrees of plantarflexion. The TAA group that was included in this experiment consisted of 11 participants, 10 of whom had received a unilateral mobile-bearing ankle arthroplasty due to post-traumatic arthritis and 1 due to rheumatoid arthritis, which manifested itself 1.9 years on average (range 0.5 to 4 years) prior to the experiment. Mean duration of ankle symptoms prior to surgery was 11.7 years (range 2.5 to 28 years). Mean age at surgery was 51.5 years (range 40 to 61 years). In seven ankles operated between 2001 and 2004 a Buechel-Pappas (BP) prosthesis (Endotec, South Orange, NJ, USA) was implanted, and in four ankles operated in 2004 a Ceramic Coated Implant (CCI) prosthesis (Van Straten Medical, Nieuwegein, The Netherlands) was used. Both prostheses are mobile-bearing designs, and thereby have no intrinsic constraints. No ankle had a deformity in the frontal plane after surgery. Dorsiflexion averaged 13.4 degrees (range 10 to 32 degrees) and plantarflexion averaged 31.8 degrees (range 24 to 45 degrees). Hindfoot motion was considerably restricted in only one patient. At the time of the experiment the LCS ankle score of the patients was a mean 93 points (range 85 to 98), and the AOFAS ankle-hindfoot score was a mean 91 points (range 85 to 100). None of the patients used walking aids or had a functional impairment of any other lower extremity joint besides the operated ankle. All patients were satisfied by the result of the TAA, and all were able to walk more than one kilometer (Table 1).

2.2. Data collection

Before the start of the experiment height and body mass of each participant were measured. Body mass was measured using the force plates in the experimental set up. In the patient group, the dorsiflexion and plantarflexion at the ankle, and pronation and supination at the hindfoot, were measured by manual goniometry.

The experiment consisted of two parts: stage one in which the mechanical work performed on the CoM (CoM work) was measured, and stage two in which the metabolic energy expenditure was determined. For each participant, the two parts of the experiment were carried out on a single day.

CoM work was measured while participants walked barefoot over a 10-m walkway, with two built-in 1×1 m custom made strain gauge force plates¹, for the measurement of the ground reaction forces. At the end of the walkway a camera unit of an optical tracking system (Optotrak, Northern Digital Inc, Waterloo, Canada) was placed, which collected the positions of an infrared marker placed on the participant's belt buckle around the waist. Both Optotrak and force plate data were synchronized and collected at a sampling rate of 500 Hz.

Before data collection, participants were given ample time to get familiar with the settings. They were allowed to walk over the walkway without explicit instruction. In the meanwhile we recorded their self-selected walking speed and observed the starting point on the walkway from which they placed at least one foot on the first force plate and two on the second force plate at their natural cadence. Walking speed was determined

¹ Resolution x,y direction 0.078 N/bit, z direction 0.159 N/bit, linearity < 1%, hysteresis < 1%, cross talk < 1%, resonance frequency 60 Hz.

automatically after each trial, using the collected position data. After each trial the participant was provided with feedback on his/her walking speed in order to pace themselves to the required speed. Walking trials were carried out at two different speeds: a self-selected walking speed (SWS, i.e., the speed they self-selected on the 10-m walkway) which allowed us to analyze walking cost at a normal daily life walking speed, and a fixed walking speed (FWS = 1.25 m/s), which allowed us to control for the effect of speed on mechanical and metabolic work. In both conditions, participants had to perform five successful trials in which their right foot was the trailing leg and five trials in which their left foot was the trailing leg. Trials were excluded when the feet did not align correctly with the force plates and when velocity of the FWS trial deviated more than 0.05 m/s from the desired 1.25 m/s or when participants clearly accelerated or decelerated during the stride over the force plates (as could be assessed during data analysis).

Subsequently, participants' metabolic energy expenditure was measured while walking on a treadmill. This was done by analyzing inspired and expired air with use of an Oxycon breath-by-breath gas analyzer (Jaeger GmbH, Hoechberg, Germany). Participants walked barefoot at the same SWS and FWS as they had adopted during the walkway trials. Before the start of treadmill walking the resting metabolism during 3 minutes of quiet standing was measured. Subsequently, participants were allowed to get accustomed to walking on a treadmill, which in general took less than 5 minutes. After this period participants walked five minutes at SWS and after a period of rest at FWS while oxygen uptake was measured. Halfway during each treadmill trial step frequency was determined in order to test whether this was similar compared to the walkway test.

2.3. Data analysis

From the walkway trials, data were analyzed for one complete step, starting with double support and ending after single support on the leading leg. Double support started with the placement of the leading leg on the second force plate and ended with the toe-off of the trailing leg on the first force plate. With this toe-off, the single support started, which ended with the placement of the former trailing leg on the second force plate. This latter instant could be found as a sudden large displacement of the center of pressure on the second force plate.

Using the force plate data, mechanical CoM work could be calculated according to the individual limbs methods outlined by Donelan, Kram, and Kuo (2002a,b). First, acceleration of the CoM was calculated from the summed ground reaction forces acting under each limb. Since velocity is the integral of acceleration, velocity of the CoM can be calculated using Eq. 1

$$\vec{v}_{com} = \int \left(\frac{\vec{F}_{trail} + \vec{F}_{lead} + m \cdot \vec{g}}{m} \right) dt + \begin{bmatrix} c_x \\ c_y \\ c_z \end{bmatrix} \quad (1)$$

where \vec{v}_{com} is the velocity vector of the CoM, \vec{F}_{trail} is the ground reaction force vector exerted by the trailing, push-off, \vec{F}_{lead} is the force exerted by the leading, new stance, m is the participants' body mass and \vec{g} is the gravitational acceleration ($[0, 0, -9.81] \text{ m}\cdot\text{s}^{-2}$). In accordance with Donelan et al. (2002a), the integration constant for the vertical direction (c_z) was obtained by assuming the average vertical CoM velocity over a step to be zero. The integration constant for the fore-aft direction (c_y) was found by assuming the average velocity over a step to be equal to the average walking speed measured by the Optotrak system. For the medio-lateral direction (c_x), the integration constant was found

by assuming the CoM velocity at the end of each step to be equal in magnitude but opposite in sign compared to the beginning.

Multiplying force under each separate limb by velocity of the body's CoM results in the calculation of the mechanical power generated by each limb on the CoM (Eqs. 2 and 3):

$$P_{trail} = \vec{F}_{trail} \cdot \vec{v}_{com} \quad (2)$$

$$P_{lead} = \vec{F}_{lead} \cdot \vec{v}_{com} \quad (3)$$

The mechanical work performed on the CoM (CoM work) is equal to the cumulative time-integral of the mechanical power and was normalized to body mass. With these calculations of mechanical work, the net mechanical work during one step was calculated for both double and single support in both trailing and leading leg. Since during steady walking the net mechanical work over a complete stride will be zero, total mechanical work over a stride was calculated by summing the absolute (negative and positive) work over two subsequent steps. To compare with metabolic parameters the mechanical work per stride is expressed as average mechanical power ($\text{W} \cdot \text{kg}^{-1}$) by dividing the mechanical work per stride (J) by body mass (kg), divided by stride time (s). The mechanical cost of transport ($\text{J} \cdot \text{kg}^{-1} \cdot \text{m}^{-1}$) is calculated by dividing the mechanical work per stride (J) by body mass (kg), divided by stride length (m). Data of five successful trials with each leg for each participant were averaged after analysis of each separate trial.

Metabolic energy consumption (\dot{E}_{met}) was calculated from VO_2 ($\text{ml} \cdot \text{s}^{-1}$) and respiratory exchange ration (RER; Garby, & Astrup, 1987)

$$\dot{E}_{met} = 4.94 \cdot RER + 16.04 \cdot \dot{\text{V}}\text{O}_2 \quad (4)$$

To derive metabolic power ($\text{W}\cdot\text{kg}^{-1}$), the metabolic energy consumption (\dot{E}_{met}) was divided by body mass (kg). Metabolic cost of transport ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$) was derived in a similar way by dividing metabolic energy consumption (\dot{E}_{met}) by body mass (kg) and divided by walking speed ($\text{m}\cdot\text{s}^{-1}$).

2.4. Statistics

The data were tested for normality using the Kolmogorov-Smirnov test, which indicated that the outcome parameters did not deviate from a normal distribution, and parametric statistics could be used. Differences between the healthy and affected leg of the TAA group and the control group were tested for significance using a Student *t*-test. For the comparison of step frequency between the walking conditions a paired sampled *t*-test was used. The correlation between mechanical work and metabolic energy cost was analyzed using Pearson's correlation coefficient. Differences were considered significant when $p < .05$.

3. Results

3.1 Spatiotemporal and metabolic parameters

Table 2 shows the spatiotemporal and metabolic parameters for the SWS and FWS condition. At SWS, control participants walked faster than the patient group. The metabolic cost of transport was 6% higher for the patient group. This difference was not statistically significant. Both controls and patients were able to walk comfortably at FWS, which was for both groups slower than their SWS. At FWS, walking was significantly

more demanding for the patient group. Metabolic power and cost of transport were respectively 29% and 28% higher for the patient group. Step length and step frequency did not differ significantly between groups at both SWS and FWS (Table 2), nor did step length and frequency differ between walking on the walkway and walking on the treadmill for both velocities.

3.2. Mechanical work

Table 3 shows the parameters dealing with the mechanical work performed on the CoM at both SWS and FWS during the separate phases of a step. For the patient group steps were analyzed separately: one step in which the healthy leg was the trailing leg (TAAh, healthy push-off) and one step in which the affected leg was the trailing leg (TAAa, affected push-off). In the two separate phases of a step, double support and single support, differences occurred, especially at the TAAa step. These differences can also be seen in the power curves of each step for SWS and FWS (Fig. 1). Note that the areas under the power curves visualize external mechanical work. Net work over a step did, on average, not exceed $0.0125 \text{ J}\cdot\text{kg}^{-1}$ for each condition.

At SWS, as hypothesized, less positive mechanical work was generated with the trailing TAA leg during the double support phase, and simultaneously more negative work was generated by the healthy leading leg compared to the control group. During the single support phase in the control group net external mechanical work was negative. For the TAA group, negative mechanical work in the single support was significantly reduced on the affected leg and was even converted into positive work in the healthy leg. The total absolute mechanical work per full step did not differ between controls and patients for

both a step with the affected and healthy leg being the trailing leg. Furthermore, the average mechanical cost of transport at SWS was significantly lower for the TAA group (see Table 2).

At FWS, less positive mechanical work was generated by the TAA trailing leg during the double support phase, and simultaneously more negative mechanical work was generated by the healthy leading leg compared to the control group. When the healthy leg was the trailing leg, during the double support phase a non-significant reduction in positive work of the healthy trailing leg was accompanied by a significant increase in the negative work in the TAA leading leg. During the single support phase, the net mechanical work was negative for controls but significantly less negative for the single support on the affected leg and even positive for the healthy leg. Differences in net mechanical work during single support were also significant between both the affected and healthy legs and the control group. Over a full step, however, total absolute mechanical work did not differ between controls and patients. Average mechanical cost of transport at FWS was not significantly reduced for the TAA group (see Table 2).

3.3. Relationship between metabolic and mechanical energy

A significant correlation was found between metabolic and total absolute mechanical cost of transport at FWS ($r = -.519$, $p = .013$; see Fig. 2). Remarkably, this correlation was negative, meaning that the more absolute mechanical work over a stride was generated the less metabolic energy was consumed. In contrast, a positive correlation was found between the negative work performed during double support (expressed as $\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$) and the metabolic cost of transport at FWS ($r = .540$, $p < .009$) and at SWS ($p = .404$, $p =$

.100), although the latter was not significant. This indicates that the larger the collision cost in the leading leg during heel strike the higher the metabolic cost of walking, i.e., the energy dissipated during the step-to-step transition explained 29% of the variance in metabolic energy cost of walking at FWS in the total group of participants.

4. Discussion

We investigated whether the impaired ankle function after TAA affected the metabolic cost of walking and the mechanical work during the step-to-step transition. At FWS, mechanical work performed by the trailing TAA leg was lower, and walking proved to be metabolically more demanding for the study group. Self-selected walking speed was 12% lower in the study group. At SWS, however, no significant difference in metabolic power or cost of transport was found.

The only study on energy expenditure after TAA we are aware of is the study by Detrembleur and Leemrijse (2009). In a prospective study, they found an increase in walking speed and a decrease in external mechanical work and energy expenditure at an average of 7 months after TAA surgery. However, in their patients walking speed remained well below ($0.77 \text{ m}\cdot\text{s}^{-1}$) and metabolic cost of transport remained well above ($3.18 \text{ J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$) the level of able bodied participants and the level of the patient group included in this study who were, on average, 1.9 years after surgery.

What makes TAA walking at FWS more energy consuming than normal walking? An explanation was expected to be found using the double inverted pendulum model of walking (Donelan et al., 2002a,b; Kuo et al., 2005). According to this model, the

following hypotheses could be formulated regarding the result of a decrease in active plantar flexion power around the TAA ankle: 1) a decrease of positive work performed during push-off by the affected leg in the double support phase, 2) an increase of negative work performed during loading of the healthy leading leg in the double support phase, 3) an increase in work performed during single support, and subsequently 4) an increase of the total external mechanical work performed, and 5) an increase in the metabolic cost of walking at a constant velocity.

As expected, at FWS TAA patients showed a decrease in positive work during the push-off with their affected leg, an increase in negative external mechanical work performed by the healthy leading leg, and an increase in the net work during the single support phase. Hence, for these separate phases of the gait cycle our hypotheses could be confirmed and indeed we found that the mechanical work dissipated in the collision of the leading leg during double support was larger in the TAA group.

In contrast to our hypothesis, however, total external mechanical work per step and per stride did not differ between patients and controls. Moreover, a weak though significant negative correlation was even found between total external mechanical and metabolic cost of walking, i.e., the higher the metabolic cost of walking, the lower the external mechanical energy proved to be. So, despite an increased mechanical energy dissipation during the step-to-step transition no increase in total mechanical work was found. It could be possible that TAA patients found a strategy to reduce total external mechanical work, despite their increased negative work at heelstrike. However, since this strategy apparently does not reduce metabolic cost, there seems no benefit in doing so. Alternatively, the relationship between total external mechanical work and metabolic

energy consumption could be questioned. It should be realized that the total external mechanical work, as calculated in this study, represents the net work performed on the body's CoM. Simultaneous opposite work terms can cancel each other out. For the relatively static double support phase this cancellation is dealt with by the individual limbs method. However, especially during the single support phase, where energy is generated and exchanged between stance and swing leg, the relation between muscle work and external work is not warranted. For instance, it is not possible to derive whether the increase in external work in the single support phase found in this study, which actually was a decrease in net negative work, is the result of an increase in positive muscle work, a reduction in negative muscle work, or a combination of both. These limitations of external work calculation have been demonstrated and discussed previously (Gitter, Czerniecki, & Weaver, 1995; Neptune, Zajac, & Kautz, 2004; van Ingen Schenau & Cavanagh, 1990)

Despite the above mentioned concerns with regard to the validity of calculations of external work (which most importantly affect the single support phase), a significant increase in negative work in the leading intact leg during double support was found for the patient group. This means that extra energy is dissipated at heelstrike, which needs to be regenerated somewhere during the step. In addition, a positive correlation was found between the negative work done by the leading leg during heel strike and the metabolic cost of walking. The correlation between the negative work at heelstrike and metabolic energy cost indicates that the work dissipated at heelstrike during FWS accounts for 29% of the variance in energy cost of walking in our study group. Hence, with some prudence,

it can still be concluded that the mechanical work required for the step-to-step transition explains at least part of the increased metabolic cost of walking after TAA.

The SWS condition was included in this study since it represents a more natural condition compared to FWS, although interpretation is more difficult due to the effect of walking speed on mechanical and metabolic cost. Self selected walking speed appeared to be almost similar to the imposed fixed walking speed in the patient group, but controls walked significantly faster in the SWS condition. It was observed that work performed by the TAA leg during the separate phases of a step at SWS was similar to the work at FWS in patients, but due to the higher walking speed of the control group, the difference in collision cost with the control group became non-significant. Nevertheless, still a moderate, though not quite significant, correlation was maintained between the negative work at heelstrike and metabolic energy cost. This supports the conclusion that the energy cost for the step-to-step transition contributes to the increased metabolic cost of walking after TAA.

The explanation of the increased energy cost of pathological gait by the inverted pendulum model is encouraging, since past (biomechanical) studies aimed at explaining the increased energy cost of pathological gait have failed. For instance, the increased energy cost of amputee's walking with a lower limb prosthesis has been shown to be unrelated to vertical CoM movement, external total CoM work (using combined limbs method), joint work or recovery index (Gitter et al., 1995, Tesio, Lanzi, & Detrembleur, 1998). Looking in isolation at collision cost during the step-to-step transition might thus be useful to investigate the energy cost of other pathological gaits. This has recently also

been demonstrated for the energy cost of walking in people after lower limb amputation (Houdijk, Pollman, Groenewold, Wiggerts, & Polomski, 2009).

However, the increased mechanical work for the step-to-step transition only seems able to account for part of the increased energy cost of walking after TAA. It therefore is reasonable to consider additional mechanisms as well. One of these mechanisms could be related to balance control. In an earlier study increased co-contraction of the lower leg muscles was found in walking after TAA, possibly in an attempt to stabilize gait (Doets et al., 2007). This co-contraction is metabolically demanding but does not contribute to external mechanical work as measured with the use of ground reaction forces. An increased effort for balance control might be an intrinsic feature of walking after TAA, but might also be enhanced by the fact that participants had to walk on a treadmill, which might be more challenging for the patients than for the able-bodied control participant. Besides the balance control issue, walking on a treadmill has been found to be mechanically similar to overground walking (van Ingen Schenau, 1980), and hence will be less likely to account for differences between mechanical energy measured on the walkway and metabolic cost measured on the treadmill. Moreover, step length, which has an effect on step-to-step transition cost (Kuo, 2002), was found not to differ between walkway and treadmill trials in this study.

Another explanation could perhaps be found in the atrophy of the lower leg muscles that results from longstanding ankle disease. In atrophic muscle tissue fiber composition has changed as type 1 muscle fibers (slow, fatigue resistant) are lost predominantly. Thus, an atrophic muscle consists of a greater deal of fast, type 2, muscle fibers, which consume more energy than slow, type 1, muscle fibers (Jones, Round, & de Haan, 2004). Thus,

persistent muscle atrophy could theoretically also account for the higher metabolic energy requirements found in this study.

With some limitations, our metabolic results can be compared with other studies in which ankle function was impaired or restricted. Waters et al. (1988), found that patients with an ankle arthrodesis had an 11% increase in oxygen cost compared to healthy controls, both walking at their SWS. In healthy participants walking with and without a below-knee plaster cast the oxygen cost at SWS was found to be 16% to 27% higher compared to unconstrained walking (Fowler et al., 1993; Waters et al., 1982). For these latter studies however, the weight of the cast and the fact that participants did not walk barefooted should be taken into account. In addition, the increase of energy expenditure of walking with an externally immobilized ankle can be mitigated by the use of an appropriate rocker bottom sole (Adamczyk, Collins, & Kuo, 2006; Vanderpool, Collins, & Kuo, 2008). With a 6% increase of metabolic cost at SWS, as found in this study, the metabolic demands for TAA participants appear favorable in comparison with participants with either an externally immobilized or a surgically fused ankle, the more so as our patient group walked at a higher SWS than the participants in the referred studies.

In conclusion, in a patient group with a well-functioning unilateral total ankle arthroplasty we have found that metabolic power and cost of transport were significantly higher compared to an able-bodied control group. This coincided with, and is partially explained by a higher negative mechanical work during the collision of the leading leg in the step-to-step transition. This indicates that after a successful TAA an impaired ankle function remains, which contributes to an increased mechanical energy dissipation during the step-to-step transition and to a reduction in walking economy.

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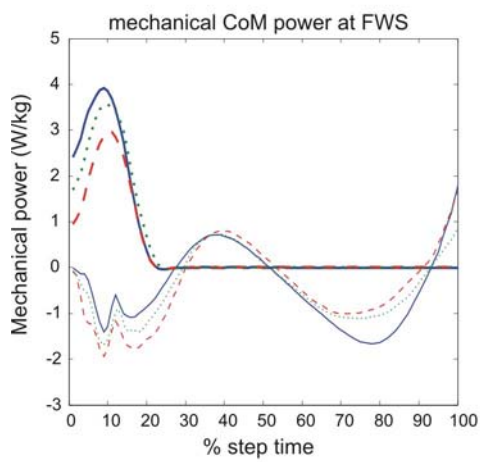
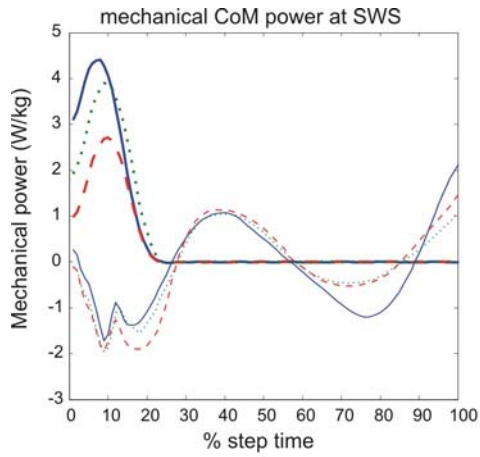
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Legends

Fig. 1. Power curves from the trailing and leading leg during walking at their Self-selected Walking Speed (SWS) and at Fixed Walking Speed (FWS) of 1.25 m s^{-1} . Areas under the power curves represent the positive and negative mechanical work. Mean curves for the control group in solid blue; mean curves for the step in which the healthy leg of patients was trailing (TAAh) in dotted green; mean curves for the step in which the affected leg of the patient was trailing (TAAa) in dashed red.

Fig. 2. Correlations between metabolic and total absolute mechanic cost of transport of TAA patients and controls. Upper panels: relation between metabolic cost and mechanical cost per stride (J kg m^{-1}) at SWS and at FWS. Lower panels: relation between metabolic cost per stride and negative mechanical work during the double support phase (J kg m^{-1}) at SWS and at FWS. Control participants are displayed as gray circles, TAA participants as blue diamonds.



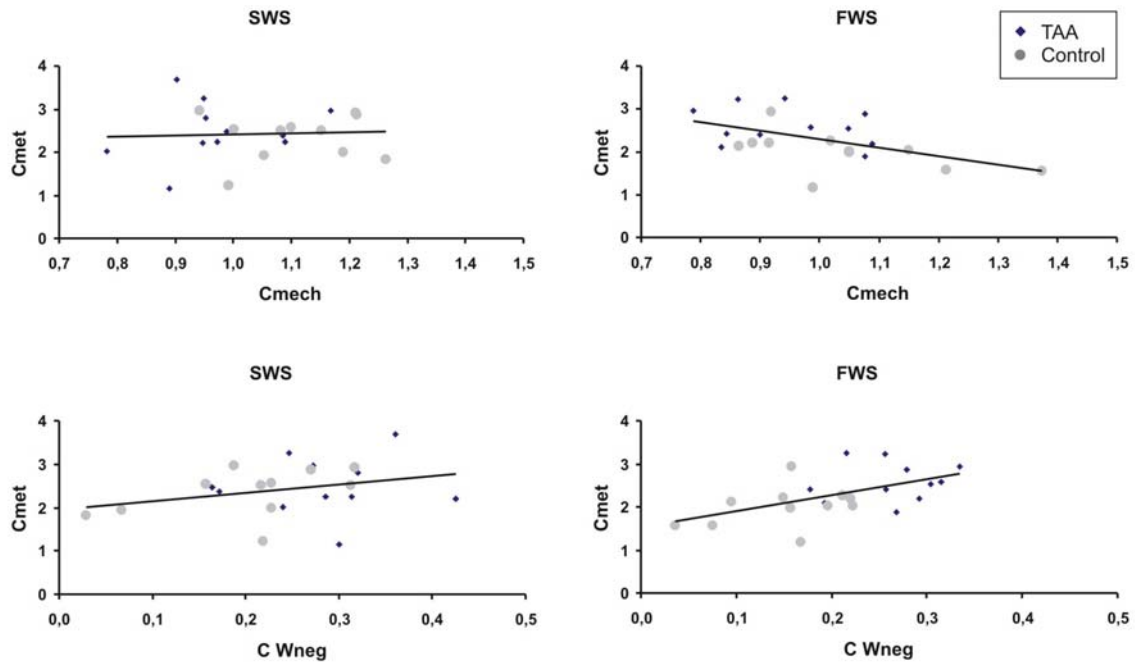


Table 1 Demographics of both study populations

	Mean, SD		<i>p</i>
	Control	TAA	
Gender	♂=7, ♀=4	♂=9, ♀=2	
Age (years)	45.4 (8.1)	54.7 (5.7)	0.005*
Mass (kg)	73.0 (8.0)	83.6 (10.3)	0.013*
Height (cm)	174.3 (7.5)	174.6 (8.4)	0.916
BMI	24.0 (2.2)	27.4 (3.1)	0.007*

TAA = Total Ankle Arthroplasty group; BMI = body mass index; * significant difference ($p < 0.05$)

Table 2 Gait parameters, metabolic power and cost of transport, and mechanical power and cost of transport for both groups during self-selected walking speed (SWS) and fixed walking speed (FWS)

	Mean (SD)		<i>p</i>
	Control	TAA	
SWS			
walking speed ($\text{m}\cdot\text{s}^{-1}$)	1.47 (0.17)	1.29 (0.14)	0.03*
step frequency ($\text{steps}\cdot\text{s}^{-1}$)	1,98 (0.16)	1,83 (0.10)	0.07
step length (m)	0.74 (0.07)	0.71 (0.10)	0.34
metabolic power ($\text{W}\cdot\text{kg}^{-1}$)	3.42 (1.0)	3.62 (1.16)	0.74
metabolic cost of transport ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$)	2.35 (0.54)	2.50 (0.68)	0.60
mechanical power ($\text{W}\cdot\text{kg}^{-1}$)	1.62 (0.28)	1.26 (0.24)	0.005*
mechanical cost of transport ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$)	1.11 (0.11)	0.98 (0.11)	0.008*
FWS			
walking speed ($\text{m}\cdot\text{s}^{-1}$)	1.25 (0.01)	1.25 (0.01)	
step frequency ($\text{steps}\cdot\text{s}^{-1}$)	1.83 (0.12)	1,81 (0.10)	0.97
step length (m)	0.69 (0.04)	0.69 (0.04)	0.99
metabolic power ($\text{W}\cdot\text{kg}^{-1}$)	2.40 (0.55)	3.09 (0.54)	0.007*
metabolic cost of transport ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$)	2.01 (0.46)	2.58 (0.45)	0.007*
mechanical power ($\text{W}\cdot\text{kg}^{-1}$)	1.30 (0.20)	1.19 (0.14)	0.14
mechanical cost of transport ($\text{J}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$)	1.04 (0.16)	0.95 (0.11)	0.14

TAA = Total Ankle Arthroplasty group; * significant difference ($p < 0.05$)

Table 3 Center of mass work (J/kg) during the separate phases of a step

	Mean (SD)			<i>p</i>	
	CO	TAAh	TAAa	CO-TAAh	CO-TAAa
SWS					
Wtrail DS	0.27 (0.05)	0.23 (0.05)	0.16 (0.04)	0.04*	< 0.001*
Wlead DS	-0.15 (0.07)	-0.18 (0.07)	-0.22 (0.09)	0.24	0.04*
W SS	-0.12 (0.09)	-0.04 (0.10)	0.06 (0.08)	0.09	< 0.001*
Wstep	0.82 (0.12)	0.67 (0.14)	0.72 (0.17)	0.013*	0.15
FWS					
Wtrail DS	0.26 (0.04)	0.22 (0.05)	0.16 (0.03)	0.061	< 0.001*
Wlead DS	-0.10 (0.06)	-0.17 (0.04)	-0.20 (0.07)	0.004*	< 0.001*
W SS	-0.15 (0.10)	-0.04 (0.05)	0.03 (0.07)	0.003*	< 0.001*
Wstep	0.72 (0.13)	0.62 (0.06)	0.69 (0.10)	0.044*	0.61

Parameters for a step in both test groups during self-selected walking speed (SWS) and fixed walking speed (FWS, 1.25 m/s): TAAh = a TAA step which started with the push-off by the healthy leg; TAAa = a TAA step which started with the push-off by the affected leg; W DS = net work in one leg (lead or trail) during double support; W SS = net work during single support; |Wstep| = absolute work over a full step; * significant difference ($p < 0.05$)